The University Of Sheffield.

Department Of Mechanical Engineering

The Development Of Test Protocols For Padded Clothing In Rugby Union Using Human Tissue Impact Surrogates

by

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ABSTRACT

Padded clothing (shoulder padding) is worn in Rugby Union to allow players to protect themselves. A performance specification for padded clothing has been set out by World Rugby™, with the intention that padding only protects against Cuts and Abrasions [1]. This performance specification is set out in Regulation 12 (padded clothing) [2]. This limits its density (45 + 15 kg·m^{-3}), thickness (10 + 2 mm) and impact attenuation performance (acceleration > 150 g for a 14.7 J impact). Regulation 12 was critiqued and areas for improvement were identified.

A literature review was conducted, covering injury and protection in Rugby Union, injury modes, anatomy of the human shoulder, organic tissue properties and human tissue simulants. It was identified that padded clothing's ability to prevent Cut and Abrasion injuries have yet to be quantitatively assessed. This was crucial in improving the Regulation 12 test protocols.

To address this problem, a multi-faceted investigation was performed. To start with, assessments were made of rugby players' external and internal shoulder anatomies using 3D and ultrasound scanning techniques. From this, geometries of rugby players' shoulders were found. The material properties of organic tissues were also assessed, with the focus being on the tissue's compressive response to load. The reason for this work was to aid the fabrication of a human shoulder surrogate.

Both a simplified and anatomical human shoulder surrogate were fabricated using human tissue simulants, as well as 3D printing and moulding techniques. A bespoke muscle simulant was developed with similar compressive properties to organic muscle tissue.

Both the simplified and anatomical surrogates were integrated into various impact testing procedures. Padded clothing was tested for its force attenuative properties, its ability to prevent blunt force Cut and Abrasion injuries, and its ability to prevent stud-induced injuries. The results from this have led to informed recommendations

being made for the improved assessment of padded clothing in Rugby Union and therefore an improved Regulation 12.

The research conducted in this thesis was the first to quantitatively assess padded rugby clothing's ability to protect from specific injuries. As well as fabricate a human shoulder surrogate for the assessment of sports padding. The testing protocols developed in this thesis can be easily adapted for the assessment of protection or padding in other collision sports or even in other industries like ballistics or automotive.

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Finally, I dedicate this Doctoral Thesis to my late Gran. You always wanted one of your grandkids to become a medical doctor, I guess this will have to do.

OUTPUTS ARISING FROM THIS RESEARCH

1. **Hughes, A.,** Driscoll, H., & Carré, M. (2020). Development of Silicone Elastomer for Use in the Assessment of Padded Clothing in Rugby Union. *In Multidisciplinary Digital Publishing Institute Proceedings*, 49(1), 77.
2. **Hughes, A.,** Driscoll, H., & Carré, M. (2021). Perceptions and Attitudes Towards Shoulder Padding and Shoulder Injury in Rugby Union. *Journal of Science in Sport and Exercise.* Accepted for publication [15/09/2021].
3. **Hughes, A.,** Dixon, J., Driscoll, H., Booth, J., & Carré, M. (2022). Padded rugby clothing to prevent laceration and abrasion injuries from stud raking: a method assessment. *Sports Engineering,* 15(1), 1-8.
4. **Hughes, A.,** Driscoll, H., & Carré, M. (2021). Development of a Simplified and Biofidelic Human Shoulder Surrogate for Testing of Padded Clothing in Rugby Union. *Abstract Submission, ESB 2021* [July 11-14, 2021].
5. **Hughes, A.,** Driscoll, H., & Carré, M. (2021). The External and Internal Geometries of Rugby Union Players' Shoulder Complex. *Abstract Submission, ISEA 2022* [accepted].
6. **Hughes, A.,** Driscoll, H., Iman, S., Allen, T., & Carré, M. (2021). Assessing the Ability of Padded Clothing to Prevent Stud Induced Injuries in Rugby Union. *Abstract Submission, ISEA 2022* [accepted].
7. Dr Tom Allen, Prof Matt Carré, Dr Heather Driscoll, **Angus Hughes** & Adil Imam. *Review of World Rugby™ Regulation 12 – Update on Report III following Manufacturers Meeting 2020*. Submitted to World Rugby™- January 2021.
8. Dr Tom Allen, Prof Matt Carré, Dr Heather Driscoll, **Angus Hughes** & Adil Imam. *Review of World Rugby™ Regulation-12 – Report III - Recommendations for Regulation-12 Update*. Submitted to World Rugby™- September 2020.
9. Dr Tom Allen, Prof Matt Carré, Dr Heather Driscoll, **Angus Hughes** & Adil Imam. *Review of World Rugby™ Regulation-12 – Padded Clothing- Report II - Update and Testbed Development.* Submitted to World Rugby™- February 2020.
10. Dr Tom Allen, Dr Matt Carré, Dr Heather Driscoll, **Angus Hughes** & Adil Imam. *Report I - Regulation-12: Critique, Assessment and Recommendations.* Submitted to World Rugby™- January 2019.

11. Dr Tom Allen, Dr Matt Carré, Dr Heather Driscoll, **Angus Hughes** & Adil Imam. *Literature Review of Injuries relevant to padded clothing used in rugby union.* Submitted to World Rugby™- April 2018.

Conferences/ Presentations

12th July 2021, Podium Presentation, ESB 2021, Development of a Simplified and Biofidelic Human Shoulder Surrogate for Testing of Padded Clothing in Rugby Union.

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17th November 2020, Oral Presentation, World Rugby™ Regulation-12 Research- World Rugby™ Manufacturers Meeting 2020.

6th July 2020, Oral Presentation, ISEA 2020, Development of Silicone Elastomer for Use in the Assessment of Padded Clothing in Rugby Union.

30th January 2020, 4th Sports Engineering Seminar Day – Sheffield Hallam University, Oral Presentation, Rugby Shoulder Padding: Advancing Impact Test Protocols.

7th July 2019, Polymer CDT Summer School, Poster Presentation, Initial development of a bespoke silicone elastomer for use as a human tissue simulant.

24th July 2018, Polymer CDT Summer School, Oral Presentation, Review of World Rugby Regulation-12.

Tables of Contents

ABBREVIATIONS

3D	Three-dimensional
AC	Acromioclavicular
ASTM	American Society for Testing and Materials
BMI	Body Mass Index
CI	Confidence Interval
CAD	Computer Aided Design
COF	Coefficient of Friction
CT	Computer Tomography
d	Displacement
FE	Finite Element
FEA	Finite Element Analysis
GH	Glenohumeral
HSV	High Speed Video
J	Joules
MMU	Manchester Metropolitan University
MRI	Magnetic Resonance Imaging
MSDS	Material Safety Data Sheet
m/s	Metres per second (Velocity)
m/s^2	Metres per second squared (Acceleration)
N	Newton's
p	Pixels
PDMS	Polydimethysiloxane
PMHS	Post-Mortem Human Subjects
PMH	Player Match Hours

PPE	Personal Protective Equipment
PPmm	Pixels per millimeter
RWC	Rugby World Cup
SHPB	Split Hopkinson Pressure Bar
UoS	University of Sheffield
UTS	Ultimate Tensile Strength
VHP	Visible Human Project
WR	World Rugby™

NOMENCLATURE

Abrasion	A skin injury as a result of rubbing.
Acceleration	Rate of change of velocity with respect to time.
Anisotropic	Exhibition of directional dependent material responses.
Anterior	Situated before or at the front of.
Biaxial	Having two axes.
Biofidelity	The closeness of surrogate to the human system it embodies.
Bovine	Relating to a cow.
Contusion	An impact injury, in which the subsurface tissue is injured but the skin is not broken, commonly known as a bruise.
Cut	A skin wound with the separation of connective tissue.
Dislocation	A joint injury whereby the ends of bones are forced from their normal positions.
Epidemiology	Study of incidence, causes ands effects of a population.
Ex vivo	That which takes place outside an organism.
Haematoma	A severe Contusion whereby a mass of clotted blood forms in a tissue or organ as a result of a broken blood vessel.
Homogeneous	Uniform in structure.
Elastic	A material capable of returning to its original shape after being stretched.
Fracture	A bone injury whereby the bone breaks.
Frontal (Coronal) Plane	A plane parallel to the long axis of the body and perpendicular to the sagittal plane that separates the body into front and back portions.
Inferior	Situated below and closer to the feet than another.
In vivo	That which takes place inside a living organism.
In vitro	That which takes place outside a living organism.
Isotropic	Having a physical property which has the same value when measured in different directions.

Kinematics	Describes the motion of an object, focusing on acceleration, speed and position
Kinetics	Describes the cause of motion.
Laceration	A deep cut or tear in the skin and underlying structures.
Lateral	Situated at or on the side or directing away from the midline.
Medial	Situated at or on the middle or directing towards the midline.
Muscle Tonicity	The elastic tension of living muscles.
Porcine	Relating to a pig.
PMHS/ Cadaver	A human corpse used as a surrogate in research.
Sagittal Plane	A longitudinal plane that divides the body of a human into left and right sections.
Stiffness	The extent to which and object resistant deformation in response to an applied load.
Superior	Situated towards the head and further away from the feet than another.
Uniaxial	Having a singular axis.
Velocity	Rate of change of position with respect to time.
Viscoelastic	The property of materials that exhibit both viscous and elastic characteristics when undergoing deformation.
§	Section sign, used for referencing individually numbered sections of a document.

CHAPTER 1 - INTRODUCTION

1.1 Chapter Overview

This chapter presents the background of the research presented in this Doctoral Thesis, it outlines the development of human surrogates for the improved assessment of padded clothing in Rugby Union. In addition, it provides rationale behind the proposed research relating this to research questions set out by the funder, World Rugby™. A project aim has been set out and further research questions complementary to World Rugby's questions are also presented. A chapter-by-chapter summary, with complementary flow diagram outlining its content and how each chapter links together is also presented.

1.2 Background

Rugby Union is a popular collision sport played by both males and females with global participation rates reaching 8.5 million in the year 2016 [3]. A broad spectrum of ages, genders, skills and fitness levels now regularly participate in the sport. The ever-developing professional era in Rugby Union has led to an increase in player size and game intensities. This results in a relatively high injury rate (79.4 per 1000 player match hours (PMH)), when compared with other contact sports such as soccer (64.4 per 1000 PMH [4]), rugby league (57.0 per 1000 PMH [5]), and martial arts (45.0 per 1000 PMH [6]).

This increase in participation, and therefore global economic impact [7] as well injury rate means the market for personal protective equipment (PPE) in rugby is like to increase. Although there are scientific studies on the effectiveness of rugby PPE, its ability to reduce injury is disputed [8]. A popular item of PPE is padded clothing, commercially known as shoulder padding. World Rugby™, Rugby Unions global governing body has set out a regulation to govern the equipment used in rugby. Specifically, regulation 12, schedule 1 (padded clothing [1]) in which the design, material specifications, performance requirements and testing methods of padded clothing are governed. World Rugby™ recognises the need to protect players with player welfare at the forefront of their 2016-2020 strategic plan. However, they also make it clear that their strategy is to limit the amount of protection used in the sport to a point where the game as we know it is not affected. Growing technological advancements in both protective materials and how they are tested create a need for the assessment of protective equipment in Rugby Union in line with World Rugby's strategic plan.

Sports PPE including padded clothing in Rugby Union is typically worn to perform one or more of the following functions: impact energy attenuation; acceleration management; load distribution and force limitation [9]. Padded rugby clothing is currently assessed in testing standards for its impact acceleration attenuation with maximum limits being set on this to not overly protect players and therefore change their behaviours.

Sports injury biomechanics research attempts to replicate 'real life' injurious scenarios so the mechanisms of injury can be understood, and injury prevention methods can be developed [10]. Human surrogates are used in injury biomechanics research to provide a representation of the living human enabling investigations to be carried out without the need for human or animal experiments. Advancing technologies have led to the development of more replicable human surrogates that facilitate this [11].

Many current safety standards used to assess sports PPE including padded clothing in Rugby Union are unrepresentative of impacts seen in the game as they use rigid anvils, unrepresentative geometries, and inaccurate testing parameters. There is therefore a need for an improved, more realistic testing approach that incorporates the use of human surrogates.

1.3 Regulation 12, Schedule 1: Specifications Relating to Players Dress, Sections 5-6

A project proposal was set out to review World Rugby's Regulation 12 (padded clothing). This regulation governs the provisions relating to padded clothing including its design, performance requirements and testing protocols. The fundamental goal was to develop and implement a new revised Regulation 12.

The research project consisted of two PhD projects that ran alongside each other. The PhD project at MMU (PhD A) looked to review the state-of-the-art in padded clothing and identify material requirements for protection against Cuts and Abrasions in Rugby Union. While doing this finite element (FE) models were developed to assess the protective properties of padded clothing in relation to Cuts and Abrasions.

The PhD project at the UoS (PhD B) looked to review the state-of-the-art in methods for testing padded clothing, investigating the best way to mechanically simulate scenarios that can lead to Cuts and Abrasions using tests with representative human tissue simulants. This work would validate PhD A's FE models.

1.4 Research Aim

The overall project aim was to create a new Regulation 12, with respect to padded clothing, to reflect the developing game of rugby.

The aim of the UoS PhD project is outlined below:

To develop appropriate test protocols for the assessment of padded clothing in Rugby Union via the use of human tissue surrogates and in turn, guide an updated and improved Regulation 12 (padded clothing).

1.5 Research Questions and Project Objectives

Four research questions were outlined in the proposal made to World Rugby™. These have in turn guided a series of individual project research questions enabling a clear structure and ability to objectively assess the success of the research.

The World Rugby™ research questions (RQs) are outlined below:

RQ 1) Is the current requirement for padded clothing appropriate for the modern game of rugby, how and why?

RQ 2) Is the current requirement for padded clothing appropriate in permitting the use of modern technology, how and why?

RQ 3) Considering that the intention for padded clothing is to continue to protect against Cuts and Abrasions only, devise an updated regulation with testing procedures that permits the latest technology.

RQ 4) If no restriction were placed on the performance of padded clothing by World Rugby™, what would the development of such clothing look like?

With these RQ's in mind, the individual UoS PhD project RQ's are outlined below:

Q1: Can bespoke human tissue simulants be fabricated that give a consistent and biofidelic response to load?

Human tissue simulants are usually selected for use through past research using 'off the shelf' best fit gelatines of elastomers. These simulants are usually developed for something that is different to the initial purpose. The development and fabrication of bespoke human tissue simulants that match selected mechanical properties of their human counterparts could improve the biofidelity of the human surrogate they make up.

Q2: Can a human shoulder surrogate with representative anatomies and geometries be fabricated feasibly in a repeatable manner?

Typical impact surrogates are unrepresentative of the human body part they embody. Differences in the surrogate's geometries and anatomies could lead to an unrealistic impact scenario. To add to this the shoulder is a complex body part. The use of state-of-the-art technologies could make the development of a human shoulder surrogate with accurate anatomies and geometries feasible. No multi-layer shoulder surrogate has been fabricated in past research.

Q3: Can a durable, biofidelic human shoulder surrogate be developed for repeatable and affordable use in a test house?

For more biofidelic surrogates to be adopted for test house use, they need to be repeatable and fabricated in an affordable manner. The development of shoulder surrogates with simplified anatomies and geometries could make this possible.

Q4: Can human shoulder surrogates be integrated into an impact testing set up in order to reconstruct specific injurious scenarios relevant to evaluations of padded rugby clothing?

Typical evaluations of padding and protection in the sports industry and beyond use unrepresentative loading conditions through the use of inaccurate test protocols and parameters. Through the systematic consideration of 'real world' impact parameters and the additions of human surrogates. Common injurious events can be reconstructed, and padded clothing's performance assessed.

With the following aims in mind, the UoS project objectives are outlined below:

Objective 1: To review current literature relating to injury mechanisms in Rugby Union, the anatomy of the human shoulder, mechanics of organic tissues, injury modes and human impact surrogates.

Objective 2: To critically review current regulations for padded clothing in rugby to identify gaps and areas of improvement.

Objective 3: To develop a dataset of human shoulder anatomies relevant to Rugby Union players.

Objective 4: To characterise the mechanical properties of organic tissues relevant to developing a shoulder impact surrogate.

Objective 5: To use the data collected in objectives 3 & 4 to develop shoulder impact surrogates relevant to assessing padded clothing in Rugby Union.

Objective 6: To assess padded clothing's protective capabilities focussing on blunt force and stud-induced injuries, with focus on Cuts and Abrasions.

Objective 7: To recommend updated test methods for the improved assessment of padded clothing in Rugby Union.

1.6 Chapter by Chapter Overview

A chapter-by-chapter overview has been presented to outline the body of work. Following this, a flow chart (Figure 1.1) illustrates the research's work plan.

Chapter 2 - Literature Review

An extensive literature review is presented, examining the injury mechanisms in Rugby Union. The anatomy of the human shoulder and its tissue properties are then examined. Relevant injury modes are reviewed, and the field of human impact surrogates is outlined.

Chapter 3 - Sports Impact Assessments and World Rugby™ Regulation 12

The key components for assessing sports impacts are reviewed and conceptual models for impacts to the shoulder devised. A critical review of the current World Rugby™, Regulation 12: padded clothing is presented. A published survey study entitled 'Perceptions and Attitudes Towards Shoulder Padding and Shoulder Injury in Rugby Union' was then used to establish the research approach taken to develop an improved Regulation 12.

Chapter 4 - Anatomical and Mechanical Assessments for The Development of a Human Shoulder Surrogate

Anatomical assessments of the human shoulder are provided, with external and internal geometrical measurements obtained both through literature and data collection on human participants. Mechanical properties of organic tissues are assessed experimentally and compared to past literature. The feasibility of simplifying these geometries for ease of fabrication and repeatability of testing is discussed. This chapter aims to inform the development of human shoulder surrogates.

Chapter 5 - Human Shoulder Surrogate Fabrication and Formulation

The development of custom-made silicones that match the selected mechanical properties of their human counterpart is detailed. The design and validation processes of both a simplified shoulder surrogate and a biofidelic shoulder surrogate for use in impact testing of padded rugby clothing are then presented.

Chapter 6 - The Assessment of Padded Clothing's Protective Capabilities

The experimental impact testing completed using both the shoulder surrogates and padded clothing is detailed. Padded clothing's protective capabilities are then assessed focussing on blunt force and stud-induced injuries. The impact testing completed was conducted under representative rugby impact conditions. Results of this are used to inform the recommendations made to World Rugby™ for new Regulation 12 testing protocols.

Chapter 7 - Recommendations in The Improved Assessment of Padded Clothing in Rugby Union

The recommendations made to World Rugby™ are outlined and backed up with scientific evidence where necessary to provide a rationale for each one.

Chapter 8 - Discussion, Conclusions and Recommendations for Future Work

The research aims and questions are discussed with reference to the project's findings. The limitations of the research are listed and potential areas for future work are highlighted. A Table highlighting why the work is novel and what it adds to the current state-of-the-art in human surrogate development and sports impact testing fields is also presented.

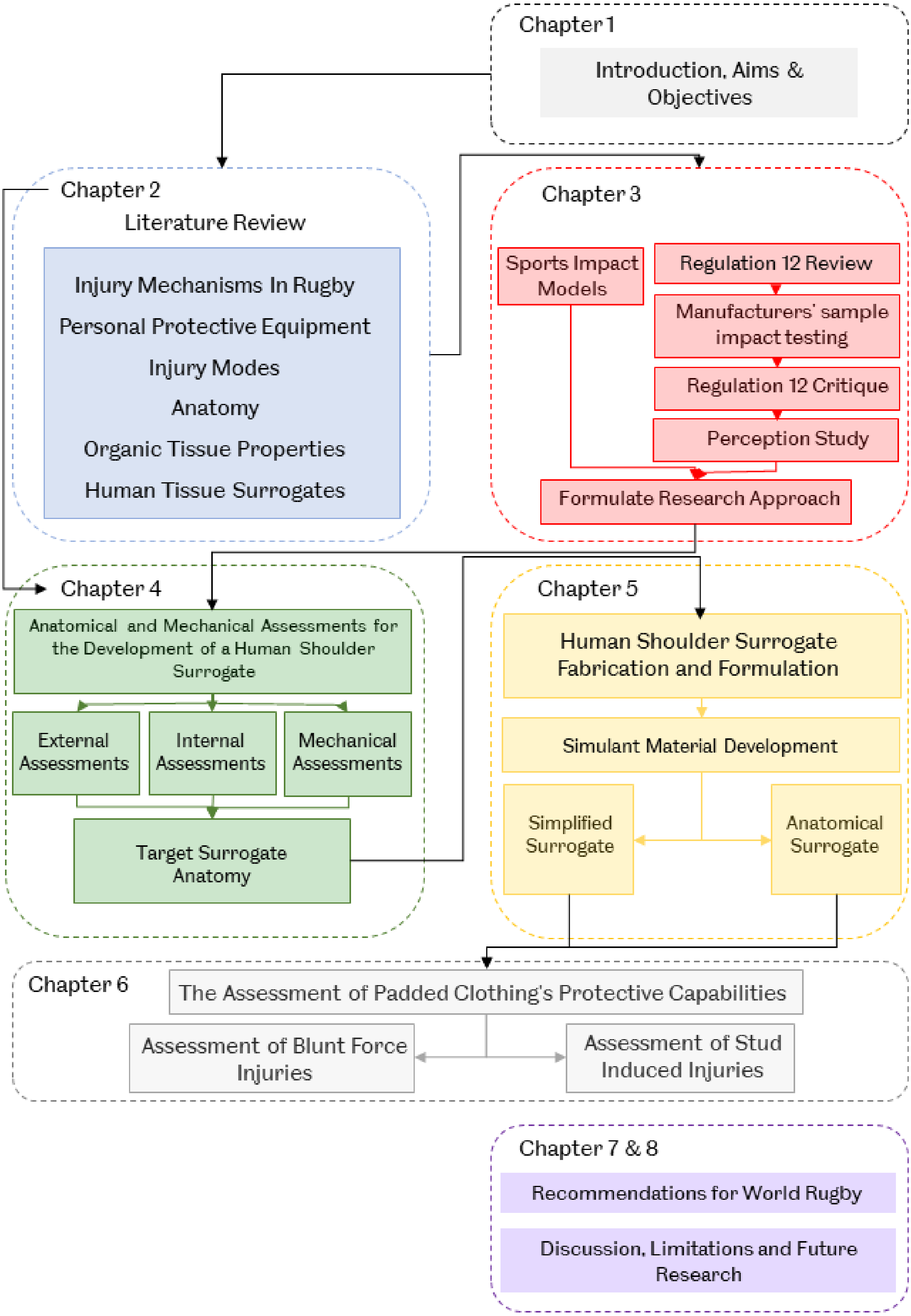


Figure 1.1 – Flow chart of project work plan.

CHAPTER 2 - LITERATURE REVIEW

2.1 Chapter Overview

This chapter provides an extensive review of the current literature concerning the injury mechanisms in Rugby Union. The causation and location of common injury modes in Rugby Union are evaluated. A review of the current methods of protection in Rugby Union has been completed. Following this, the injury modes highlighted by World Rugby™ have been reviewed. The anatomy of the shoulder was investigated, and key organic tissue structures were identified to provide an overview of their function and mechanical response. Finally, the state-of-the-art in human tissue surrogates has been outlined with consideration to developing a human shoulder surrogate.

2.2 Injury Mechanisms in Rugby Union

2.1.1 Introduction

To assess and better understand the use of personal protective equipment (PPE) in sport, the mechanisms of injury must be explored [12]. World Rugby™ has outlined the use of padded equipment as a means of reducing the severity and frequency of injuries from impacts with other players and the playing surface. It should be stressed that World Rugby™ only extends this definition to the prevention of Cuts and Abrasions. It can also be argued that protective clothing in rugby can be used as a means of impact attenuation and therefore prevent the chance of any impact injury [13]. Injuries in sport are found to be regular, although the trauma is found to be far less severe when compared with car crash or ballistic injures for example. It is suggested that this is because athletes can experience multiple loads to the same part of the body during competition or training [14]. In order to report injury, definitions must be made. This becomes more complex as definitions vary within and between sports as well as researchers [15]. Clarsen & Bahr [16] suggested 'one size does not fit all' when it comes to injury definition and the definition chosen in research should reflect the aims and context of the surveillance. Various definitions can be seen in Table 2.1. Some definitions use the mechanics that cause injury as a description, whereas some describe the events an athlete may take after being injured as a form of explanation. In relation to research, this lack of clarity for a definitive sports injury definition could lead to conflicting conclusions and recommendations reducing the value of many epidemiological studies of sports injury [17].

Table 2.1 – Sports injury definition.

Author	Definition
Timpka et al. [18]	Loss or abnormality of bodily structure or functioning resulting from an isolated exposure to physical energy during sports training or competition that following examination is diagnosed by a clinical professional as a medically recognized injury.
McIntosh [19]	A transfer of energy to the tissues that exceeds the ability to maintain structural and/ or functional integrity.
Van Mechelen et al. [20]	Limits athletic participation for at least the day after the day of the onset.
Clarsen et al. [21]	Injuries leading to the athlete seeking attention from a qualified medical practitioner.
	Rugby Specific
Fuller et al. [22]	Any physical complaint, which was caused by a transfer of energy that exceeded the body's ability to maintain its structural and/or functional integrity, that was sustained by a player during a rugby match or rugby training, irrespective of the need for medical attention or time-loss from rugby activities. An injury that results in a player receiving medical attention is referred to as a 'medical-attention' injury and an injury that results in a player being unable to take a full part in future rugby training or match play as a 'time-loss' injury.
Brooks et al. [23]	Any injury that prevents a player from taking a full part in all training and match play activities typically planned for that day for a period of greater than 24 hours from midnight at the end of the day the injury was sustained.
Best et al. [24]	Any injury or medical condition related to a game event that caused a player to leave the field during a game and/or to miss a subsequent game.

It is important definitions of injuries specifically in Rugby Union are discussed. The definition by Fuller et al. [22] is one approved by World Rugby™. This definition comes in two parts. A 'medical-attention' injury and a 'time-loss' injury. Previous research in Rugby Union has focused on the definition as something that causes time loss to a player's normal match or training regime [25-27]. Injures that cause time loss will therefore be of a severity that rugby protection may not have the ability to prevent. This causes issues in the assessment of protective clothing, as injures not causing time loss prevented by protective clothing are not included. To develop a broader understanding of injury, a definition that encompasses the true rate of incidence and severity is needed. The varying sports injury definitions can therefore distort injury statistics as well as obscure the mechanisms that may be causing injury in the chosen sport. This can therefore complicate research as true injury statistics are unknown. This is seen in Rugby Union when reporting Laceration, Abrasion and Contusion (bruises) injury. Oudshoorn et al. [28] have suggested current injury definitions within Rugby Union can lead to an underestimation of skin and Laceration injury prevalence. The need for protection to prevent more minor injuries may then be overlooked with the performance of rugby protection being over-estimated. It should be noted that a Laceration injury is generally a type of Cut, caused by blunt force impact on soft tissue. In epidemiological studies, a Laceration is generally recorded, this is explained further in §2.3.2. There is a need for future studies to focus on injuries in rugby that may not cause time loss but modify bodily structure or functioning as used by Timpka et al. [18].

Sports injuries can be classified as acute; these injuries occur due to sudden trauma to the tissue with symptoms occurring almost immediately. Sports injuries can also be classified as overuse injuries. These can occur over a period of time due to excessive and repetitive loading of tissue. Acute and overuse injuries can be arranged into the different sites shown in Figure 2.1.

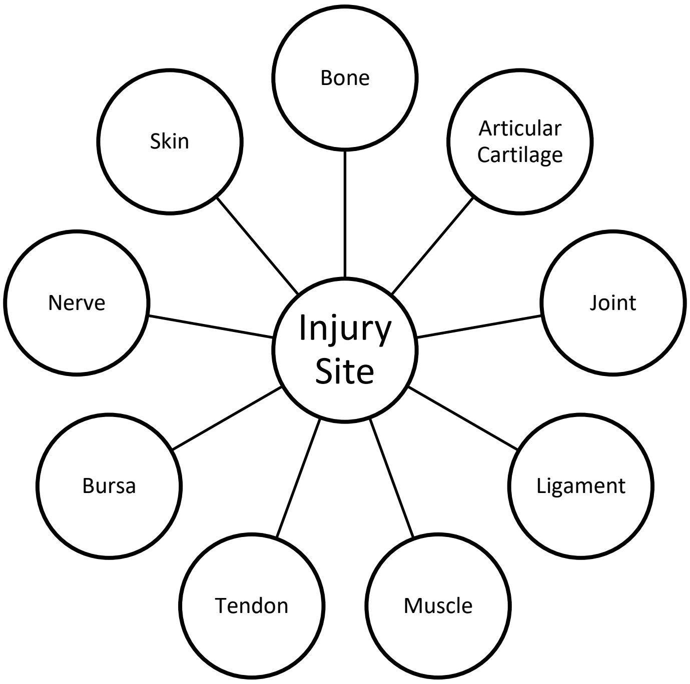


Figure 2.1 – Classification of injury sites (adapted from Brukner [29]).

The aim of this review was therefore to assess the current literature relating to injury in Rugby Union, to establish the injury mechanisms in the modern game. From this, a clear understanding can be created into how an injury is caused as well as how protection can prevent and reduce the severity of the injury.

2.1.2 Injury Statistics

Overview

Rugby union is a collision sport, resulting in a relatively high injury rate of 90.1 per 1,000 player match hours (PMH) in elite rugby [4], this however is lower in the amateur game with 46.8 injuries per 1000 PMH [30]. This can be compared to soccer with 64.4 injuries per 1000 match hours or tennis with 31.1 injuries per 1000 match hours [31]. With World Rugby™ [32] proposing that padded clothing should protect from Cuts and Abrasions, it is important to focus this review on these types of injuries. Less severe impact injuries like Contusions have also been explored.

Gerrard [33] suggests shoulder padding may protect from Cuts as well as reduce bruising of the soft tissue surrounding the shoulder. This is further backed up by Pain et al. [34] who found shoulder padding can extend the duration and broaden the area of impact so the wearer experiences decreased impact forces at contact. Padded clothing, however, is designed not to limit the range of movement at the shoulder, coupled with this, few designs protect the lateral shoulder. They, therefore, do not have the ability to prevent more severe injuries like dislocations or fractures [13]. It is, therefore, necessary to understand the nature of Laceration, Abrasion and Contusion injuries to update regulations regarding padding clothing.

Lacerations, Abrasions and Contusions

Lacerations, Abrasions and Contusions may be considered less severe than many other rugby injuries. Brooks et al. [25] found Laceration and skin injuries to cause on average 5.3 days of time loss. Fuller et al. [35] found Contusion injuries to cause on average 5.0 days of time loss. If compared to joint and ligament injury (20.6 days) or bone stress and fractures (50.2 days), this inferior time loss suggests these types of injuries to be far less severe. However, this time loss in a players' normal training regime will still cause a loss in performance and with regards to the professional game, a financial loss.

Table 2.2 summarises the Laceration, Abrasion and Contusion rates found in sixteen previous research articles, eleven of these measured injury rates at an elite level, three at amateur level and one at all levels. Laceration injury rates per 1000 player hours range from 0 - 21.24 (0 – 27 % total injuries), Abrasions 0 - 1.57 (0 - 1.6 % total injuries) and Contusions 0.09 – 19.68 (4.5 – 24 % total injuries). This range in results is seen because of the differing injury definitions. Estimating the true rate of these specific injuries then becomes problematic.

Best et al. [24] used the definition as stated in Table 2.1, 21.24 Lacerations were seen per 1000 player hours with this making up 21.7% of the total injuries. Whereas, Fuller et al. [36] used the definition referring time loss over one day as stated by Fuller et al. (2007) in Table 2.1. Only 1.6 Laceration injuries per 1000 player hours were recorded making up 1.8 % of injuries. This definition has been used to define rugby injuries since 2005 potentially distorting the prevalence of Laceration, Abrasion and Contusion injuries in the modern game. Future research must look into both the injury definition and the reporting of these injuries.

The ever-increasing use of artificial pitches in Rugby Union has led to increased skin-related injuries. The RFU plans to build 60 artificial pitches in the next four years [37], therefore the effects this playing surface has on injuries must be reviewed. Williams [38] completed a study at elite level. It was found Abrasion injuries occur 119 times every 1000 player hours on artificial turf. The four locations where an Abrasion occurred were the knee (74 %), followed by the lower leg (9 %), elbow (7 %), and forearm (4 %). It is important to note that an Abrasion did not have to induce a time loss to be reported as an injury, therefore resulting in the high incidence rate when compared with the incidence rates found and summarised in Table 2.2. It was also found Abrasions occur 15 times every 1000 player hours on natural turf. Suggesting previous literature has underestimated Abrasion injury.

Table 2.2 – Laceration, Abrasion and Contusion injury statistics.

Study	Playing Level	Injury Definition	Total Recorded Injuries	Lacerations		Abrasions		Contusions/ Haematomas	
				Overall Incidence Rate	% Total Injuries	Overall Incidence Rate	% Total Injuries	Overall Incidence Rate	% Total Injuries
[39]	International – RWC 1998	Leave the field	70 M	8.64	27.00	-	-	7.68	24.00 (All muscle injuries)
[40]	International – Australia	Leave the field or miss subsequent game	143 M	15.87	23.00	-	-	6.90	10.00
[41]	School Boy – New Zealand	Physical Complaint	340 M+T	1.45	2.90	-	-	10.60	21.50
[24]	International – RWC 2003	Leave the field or miss subsequent game	189 M	21.24	21.70	1.57	1.60	19.68	20.1
[25]	English Premiership	Time loss	1534 M	1.55	1.17	-	-	16.30	17.91
[26]	English Premiership	Time loss	395 T	0.02	1.01	-	-	0.09	4.50
[42]	International – England	Time loss	145 M+T	0.23	1.38	-	-	2.46	14.48
[43]	High School - USA	Time loss & medical attention	594 M+T	-	-	-	-	0.47	9.10
[35]	International – RWC 2007	Time loss 24hrs +	161 M	0	0	0	0	14.60	17.40
[44]	Collegiate Rugby - USA	Time loss 24hrs +	582 M	0.94	5.50	-	-	-	-
			265 T	0.25	4.53	-	-	-	-
[45]	Elite – Super 14	Time loss 24hrs +	345 M	3.95	4.10	-	-	-	-
[36]	International – RWC 2011	Time loss 24hrs +	171 M	1.60	1.80	0	0	16.10	18.07
[4]	International – RWC 2015	Time loss 24hrs +	173 M	1.53	1.70	0	0	9.91	11.00
[46]	International – RWC 2019	Time loss 24hrs +	143 M	2.22	2.8			10.56	13.30
[47]	Elite – Super 15	Time loss 24hrs +	160 M+T	6.00	4.00	-	-	-	-
[48]	Level 1 – Elite – USA	Athlete went to hospital	128,813	-	-	-	-	-	14.40

M is match injuries; T is training injuries
Overall incidence rate is reported player 1000 player hours

2.1.3 Injury Causation

Overview

Rugby Union includes four main phases of play. The tackle, ruck and maul, set-piece (scrum and lineout), and open play. The main cause of injury is the tackle (36 % – 56.3 %) [23, 24, 36, 42, 47]. It can be argued that injury rates are inclined to be high with the tackle occurring in half of all contact events in the game (221 per game) [49]. Brooks et al. [42] showed that the injury incidence rates of the player being tackled are double that of the tackler (26 % - 13 %). On the contrary Best et al., (2005) and Whitehouse et al., (2016) found the tackler and the tackled player to have similar injury incidence rates (Table 2.3). It is therefore important that the mechanisms that may cause injury in the tackle situation are explored.

Injury incidence during the ruck and maul are widespread from study to study (5.4 % – 22.8 %). The ruck is the second most frequent contact event in rugby with 142.5 incidences per game, therefore a high injury rate is expected [49]. More recent studies find a lower injury incidence rate than older studies. This may be due to more stringent law updates regarding the ruck and maul to make these contact situations safer where it is illegal to stamp on a player. More stringent laws have also been put on the tackler not 'rolling away'. This meaning the tackler on the floor will have to move away from an unsafe position faster. Overall, the ruck has now become a less contested area in the modern game with defensive teams competing for the ball far less frequently meaning injury-causing impacts are reduced [50].

A high proportion of injuries are also seen within a collision. Many injury surveillance reports do not define what a collision in rugby is making it difficult to draw conclusions from the research. However, Fuller et al. [49] defines a collision as the event in which the 'tackler attempts to stop the ball-carrier without the use of his arm(s)'. Fuller found that although far less frequent, a collision was 70% more likely to cause an injury than a tackle. It can be suggested this is because, no use of the arms in the tackle is considered the wrong technique as well as being illegal, therefore becoming a more dangerous method of making a tackle.

Table 2.3 – Incidence of injury as function of match event.

Match activity	Proportion (%)						
	RWC 2003 [24]	RWC 2007 [35]	RWC 2011 [36]	RWC 2015 [4]	RWC 2019 [46]	English Premiership (Brooks et al., 2005a)	Super Rugby [51]
Tackled	19.1	26.7	24.6	24.7	19.1	35.2	22.5
Tackling	20.6	8.1	20.4	21.2	28.7	21.1	20.7
Ruck/ Maul	5.8	18.0	7.5	9.4	7.3	22.8	5.4
Collision	16.4	17.4	11.1	17.1	16.9	11.9	19.8
Scrum	5.3	5.0	2.9	2.9	3.7	4.3	2.7
Lineout	1.1	0.6	0.6	1.8	0	0.8	0.9
Other	31.7	16.7	20.7	2.9	5.9	5.9	12.7
Not identified	-	7.5	12.2	-	-	-	15.3

Further World Rugby™ rule changes introduced to the 2017/2018 rugby season have increased contact involvements within the game. An RFU game trends summary data report found tackles to be up 11.4 % and rucks up 11.9 % [52]. Further injury surveillance must be conducted throughout the season to link injury with these contact events.

The Tackle

As mentioned, the tackle is the most common cause of rugby injury, with injuries occurring throughout the body for both carriers and tacklers. Injuries within the tackle are mainly caused my impacts between players (Carrier 45 %, Tackler 69 %) but also by impact with the field of play (Carrier 19 %, Tackler 12 %) as well as the loading a tackle may put on a player (Carrier 29 %, Tackler 9 %) [53].

It is important to consider the tackle events that cause injury. Quarrie and Hopkins [53] assessed 87494 tackle events in professional rugby. Injury assessments to ball carriers occurred at a rate of 7.0 per 1000 tackles with assessments to tacklers occurring at a rate of 5.2 per 1000 tackles. Time loss injuries to ball carriers were seen most frequently as a result of high (chest area) or middle (below chest to knees) tackles from the front or the side, 33 injuries from high tackles and 32 from middle tackles were seen based on 81 injuries from 25 587 tackle events with a single tackler only. However, this is because of the increased frequency in which these tackle heights take place. Ball carriers were at the highest risk of injury from tackles to the head-neck region (8.5 injuries per 1000 head-neck tackles), while tacklers were most at risk when making low tackles (2.2 injuries per 1000 low tackles). Fuller et al. [54] completed a similar study on English premiership matches over two seasons. 244 tackle injuries were recorded. Similar results were found with increased chance of injury seen in tackles to the head/ neck region with a 10.5 % chance of injury when the ball carrier or tackler is struck in the head/ neck. As expected, higher speed of tackles increased the chance of injury, high speed tackles resulted in a 3.2% chance of injury with low speed and stationary tackles incurring a 1.3 % and 1.2 % chance respectively.

Whether a tackle in rugby causes injury depends on four factors; the amount of energy transferred, the size of the area in which the force is distributed, the direction of the force, and the biomechanical properties of the body structures to which the energy is transferred to and from. Tackles in a frontal direction tend to cause more injuries to both the carrier and the tackler than any other direction. This is because the energy of the carrier and tackler is acting in opposite directions upon impact [55].

The amount of energy transferred within a tackle will influence injury, it is therefore important to review the velocities, accelerations and forces within a tackle and how these may affect injury. Average velocities just before a front-on tackle in professional rugby are 4.8 $m.s^{-1}$ (ball carrier) and 5.6 $m.s^{-1}$ (tackler). This is similar for side-on tackles (4.7 $m.s^{-1}$ ball carrier, 5.2 $m.s^{-1}$ tackler). Average accelerations 0.5 seconds before contact show a players slow down before contact in both a front on (-1.24 $m.s^{-2}$ ball carrier, -1.62 $m.s^{-2}$ tackler) and side-on tackle (-1.26 $m.s^{-2}$ ball carrier, -2.44 $m.s^{-2}$ tackler) [56]. How these velocities and accelerations can influence injury is yet to be explored. The use of a two-dimensional system to measure velocities and accelerations in this study must however be questioned.

An investigation into the effects of shoulder padding on peak forces at the shoulder shows that tackle forces can range massively (472 – 2789 N without padding, 472 – 1679 N with padding) [57]. There are few relevant human injury tolerance data studies for the shoulder complex. Studies of lateral impacts to the shoulder indicate that the injury risk increases when forces exceed 3430 N [58]. This would therefore suggest this study failed to reach impact forces that may cause shoulder injury. This is because of a small mean weight of the participants (83 kg) when compared to professional participants (99.2 kg) [59], as well as the in vivo test methodology which decreases the ecological validity of the study. Payne et al., [11] summarise the conditions experienced in an 'in play' frontal rugby tackle to the thigh region of the ball carrier shown in Table 2.4. Identification of the specific conditions needed to cause injury must be developed.

Usman et al. [60] completed a study on active shoulder tackle forces in a laboratory setting. Mean peak forces of 1708 N and 1610 N were seen on the dominant and non-dominant shoulders. As with other studies, the participant's mean body mass (85 kg) was far below that expected in a professional game (99.3 kg). Seminati et al. [61] completed a similar laboratory study, however, a population with similar body mass to professional rugby players was used (96.6 kg). Peak mean impact forces ranged from 2840 N in a stationary tackle to 3400 N in a dynamic tackle. A further study by Seminati et al. [62] exhibited higher impact forces of 5300 ± 1000 N. This was in the frontal direction on the tackler's dominant side. There is a large range of rugby tackle loads dependent on their biomechanics and intensities. Mean contact time was also recorded. Mean stationary tackle contact time ranged from 0.102 to 0.111s between dominant and non-dominant sides, as well as mean contact time being 0.095s in the dynamic conditions. These larger tackle forces may be closer to the forces seen on the pitch in the professional game. Currently, there is no published 'in-play' biomechanical data that considers tackle impact forces and energies. This data could direct injury prevention research in the future.

Table 2.4 – Conditions experienced from a Rugby Union shoulder tackle (adapted from Payne et al. [11]

Scenario	Description	Impact Characteristics			
		Impactor mass (kg)	Impact velocity ($m.s^{-1}$)	Estimation of max momentum ($kg.ms^{-1}$)	Estimation of max impact energy (J)
Rugby shoulder tackle	Frontal impact with shoulder of the tackler into the ball carriers thigh	100 -230 (1-2.3* body weight assuming 100kg player)	5.6 tackler, 7.5 ball carrier	2.42×10^3	1.27×10^4

25 – 75 % of injuries to the tackler occur at the shoulder, making it the most common injury site [49, 53, 54]. Another study found that 40 % of all time-loss shoulder injuries occur to the tackler within a tackle [63]. It is therefore important to assess the impact characteristics that may cause these injuries.

It is vital to link protective clothing's purpose of preventing Cuts and Abrasions as well as providing impact attenuation to the injury mechanisms in a tackle. Prevalent injuries seen in a tackle are thigh and shoulder Haematomas as well as AC joint injuries, these are caused by impact or rotational forces [64]. Previous research has found no Cuts and Abrasion injuries to be caused by the rugby tackle [53, 65]. Protective clothing may be seen to be providing impact attenuation rather than as means of protecting against Cuts and Abrasions. Contrary to this a study by Oushoorn et al., [66] found the tackle to be the second-highest (16 %) causation of a stud injury. This is caused by a player tackling from behind and the ball carrier's studs causing injury to the tackler's upper body region. These conflicting results are caused by differences in injury definition.

The Ruck and Studs

As mentioned before the ruck is a frequent contact event in rugby where both teams can compete for the ball after a tackle. Players use their shoulders at high velocities to clear an opposition player who may be moving or stationary off the ball. The ball carrier and tackler may also be left exposed on the floor to the clearing players.

Fuller et al. [49] found the top two injuries caused by the ruck to be calf Haematoma and ankle ligament damage. It can be assumed this will be caused by both impact and rotational forces between players. As well as player-on-player interactions, the ruck is an area of the game where stud player interactions become prevalent. Oudshoorn et al. [66] found the ruck to be the causation of 56% of stud injuries. It was also found that 35% of these injuries were caused by stamping from an opposition player. It is therefore clear the ruck is a large cause of Laceration injures caused by the stud. However, additional research needs to be directed to the body location of stud injuries to build knowledge of how protective clothing can prevent Cuts and Abrasions caused by player's studs in the ruck.

Limited research has observed the kinetic and kinematic parameters of the stud impacts at the ruck. However, a recent study investigated the mechanics of stud interactions when a player stamps on a ball carrier lying down during the ruck [67]. Results of this study are seen in Table 2.5.

Table 2.5 – Kinetics and Kinematics of stamping at ruck (adapted from Oudshoorn et al. [67])

Foot inbound velocity ($m.s^{-1}$)	Foot inbound angle (°)	Peak total force (N)	Total effective mass (kg)	Total impact energy (J)	Peak stud force (N)	Stud effective mass (kg)	Stud impact energy (J)
4.3	10.3	1245.5	6.5	56.9	214.0	1.2	9.5

**All results are reported as mean of all trials.*

A limitation to this study was the low average mass of the participants (76 kg) when compared to the professional population (99.2 kg) [59]. Coupled with this, the testing was in-vivo therefore true peak forces and impact energies may have been underestimated. However, the total impact energy of 56.9 J is far greater than the current regulation for testing padded clothing where an impact energy of 14.7 J is used. This study must be compared with literature that quantifies the mechanisms that will damage human tissue to draw conclusions on the injury risk stamping in a ruck can cause.

2.1.4 Injury Location

To determine the mechanics regarding injury in Rugby Union as well as the need for protection it is important to understand the location in which injuries occur. The nature of rugby means injuries can occur to any location on the body. Understanding the frequency of these injury locations can inform the use of preventative techniques like wearing protection. Figure 2.2 summarises the frequency of injury regarding location as a percentage. The location with the most prevalent injury rate is the head and neck region (22.2%), although this result can differ due to the different injury definitions used. Many of these injuries are superficial Laceration and soft tissue injuries [27]. Protection to this region of the body to decrease the risk of head injury is therefore warranted. 10.8% of injuries occur in the shoulder and upper arm region. This would warrant the use of shoulder protection as an injury preventative method. Commercially available protective clothing also has built-in padding to the sternum and rib regions of the body. This is justified with 10.2% of injuries occurring to the trunk region. Figure 2.2 shows that injury can occur to any location of the body with its prevalence being spread through the whole body.

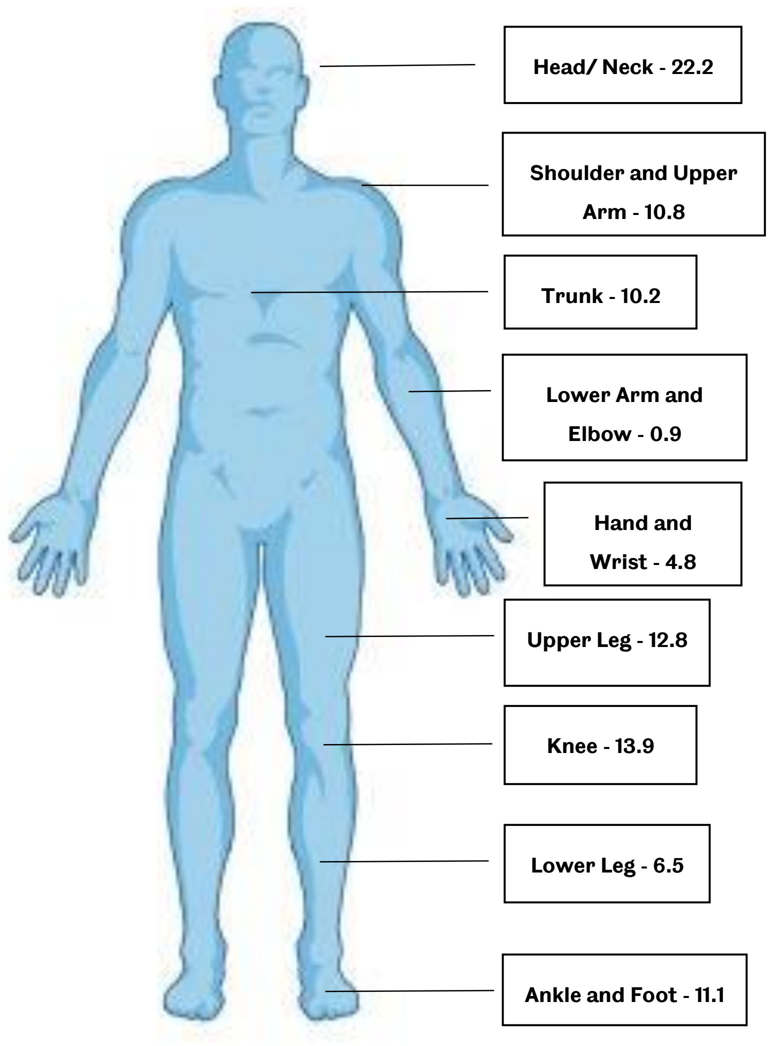


Figure 2.2 – Average injury rates by body region as a percentage (%) taken from injury surveillance reports [24, 35, 36, 68].

Shoulder injuries

To direct the development of injury prevention methods like shoulder pads it's important to understand the incidence, patterns, and mechanisms of shoulder injury in Rugby Union. Few studies have focussed solely on shoulder injuries in Rugby Union despite their frequency and severity. Namely, Usman et al. [64] completed a study on shoulder injuries in professional southern hemisphere Rugby Union. 100 shoulder injuries were recorded, the three most common injuries were Acromioclavicular (AC) joint injuries (29 %), Haematomas (17 %) and Dislocations (14 %). 77 % of shoulder injuries occurred in the tackle, 11 % in open play, 8 % in a ruck/maul and 4 % in lineouts and scrums. Usman highlighted the primary mechanisms for shoulder AC joint injury. It was underlined that the first point of contact between the tackler's elbow was the ground with the shoulder/ arm in abduction with the elbow flexed.

The high impact force with the elbow and the ground is transmitted up the arm to the shoulder causing an AC joint sprain.

Headey et al. [63] completed a similar epidemiological study on the English premiership during two seasons between 2002 and 2004. 149 injuries were recorded, and similar findings were seen in Usman's study. AC joint injuries were the most common (32 %), followed by rotator cuff/ shoulder impingement (23 %), Dislocations (14 %) and Haematomas (12 %). Helgeson and Stonemans [69] article highlighted the mechanisms of shoulder injuries in rugby players, five separate mechanisms to shoulder injury in the tackle were suggested and summarised in Table 2.6

Table 2.6 – Key mechanisms to cause shoulder injury in the Rugby Union tackle.

Affected player	Description
Tackler	Tackle with just the arm creates a leveraging force across the anterior shoulder
Tackler	The tackler encounters an impact force to the anterior shoulder as well as a leveraging force across the shoulder
Tackler and ball carrier	Impact forces are sustained to both the tackler and the ball carrier
Tackler and ball carrier	During a smothering tackle, the tackler attempts to hold or rip the ball
Ball carrier	The ball carrier is brought to the ground and impact force travels up the arm.

Although there is a substantial amount of research regarding the location of rugby injury, there is little to no research on the mechanics of injury at each body location. Future research should focus on the causation events of injury to specific locations to direct the development of protective clothing in rugby.

2.1.5 Summary

Many research points can be taken from this section. It is important to consider the definition of injury when defining the Injury mechanisms that occur in rugby. A clear definition is needed to prevent complications in research. Although there is research that quantifies the mechanics of injury-causing events in rugby, there is no research that sets the mechanical limits that would have to be achieved to cause these injuries in rugby. Literature relating to the mechanics that can cause injury to the shoulder region must therefore be examined. Specific mechanisms that are likely to cause injury in rugby can then be set.

2.2 Personal Protective Equipment in Rugby Union

Sports personal protective equipment (PPE) is worn worldwide. Its primary purpose is to reduce the risk of injury and decrease the severity of the injuries sustained. With the cost of injury continually increasing to sport, the PPE industry has become a lucrative business in which is it valued at $1.9 billion just in North America [70]. Sports manufacturers, therefore, input large amounts of resources into the research and development of sports PPE.

In Rugby Union, PPE can be split into many types; headgear, padded clothing (shoulder pads) breast padding, and mouthguard are examples. WR intends to allow for padded clothing to protect against Cut and Abrasion injuries only. However, the force attenuation capabilities of shoulder pads may also protect from other impact injuries like Contusions [55]. The amount of commercially available shoulder padding has increased year on year with World Rugby™ approved products now reaching 239. World Rugby™ has set specific requirements regarding shoulder padding. The padded material must have a density no greater than 45 kg/m^3 and an impact attenuation requirement of the peak acceleration of impacts delivered in testing exceeding 150 g. Therefore, there is a limit to the impact attenuation the padding can have. This is to ensure padding does not influence the modern game in a way that will disrupt play or overprotect players.

Many materials can be used as means of impact protection. Dilatants and polymer-based materials are commonly used for impact attenuation. These are shear-thickening materials where viscosity by shear applied strain. At high rates of deformation, the material undergoes a substantial increase in viscosity and therefore becomes stiff or rigid [71]. Modern shoulder protection regularly uses Ethylene vinyl acetate (EVA) foams as means of impact attenuation. They are a specific type of cross-linked closed-cell polyethylene foam that is soft with a rubber-like texture whilst also showing good shape recovery after deformation. Table 2.7 summarises the shoulder protection commercially available from leading brands.

Hayes and Venkatraman [55] also suggested some commercially available impact-resistant materials that could be used in protective clothing. These included Poron XRD (open-cell urethane foam), D3O (Polyurethane energy-absorbing material containing several additives and Polyborodimethylsiloxane) as well as Sorbothane. Although all materials are excellent impact attenuators, the flexibility of EVA foam lends itself to use as shoulder protection.

Few studies have investigated the effectiveness of shoulder padding, with most questioning their ability to prevent injury. Harris and Spears [13] completed an in vitro study on shoulder padding examining their force attenuation capabilities. The shoulder pads were placed on a force plate, and impactors dropped onto them, at 500, 1000 and 1500 N, a 2 kg medicine ball and 1kg hockey ball were used as the impactors to model low and high stiffness impacts. Peak force attenuation varied from 1 – 70 %, the paddings attenuation was smaller under low-stiffness impacts. Under the high stiffness impacts, the pad's impact attenuation ability varied, providing increased attenuation under low loads but decreased attenuation under higher loads. Although shoulder padding acts as an effective method of force attenuation, its ability to prevent injury can still be questioned.

Pain et al. [34] investigated the effect of rugby shoulder padding in vivo, Tekscan sensors were fitted to shoulder padding as well as the participant's shoulders and a tackle situation replicated with and without padding. A 40% decrease in peak force was seen when wearing padding, although their ability to decrease force to the AC joint area was limited. It was concluded that although their impact attenuation abilities were clear there was no evidence to suggest shoulder padding leads to a decrease in the chance of injury. Contrary to the results from the above studies, Usman et al. [60] investigated shoulder forces in the tackle with and without shoulder pads. Shoulder pads did not reduce the peak forces applied to the shoulder significantly. Shoulder padding's impact attenuation abilities must be studied further in order to direct the present study as well as establish their injury prevention capabilities.

Table 2.7- Protective clothing in rugby (leading brands).

Brand	Picture of Product	Description
Canterbury (www.canterbury.com)		EVA hexagonal foam on shoulders with EVA plate foam on chest, bicep and kidney providing flexibility and protection.
Gilbert (www.gilbertrugby.com)		Material is undisclosed by the manufacturer. Padding covers shoulders, Biceps, Sternum, Ribs, Kidneys and back.
Optimum (www.optimumsport.com)		10mm thick EVA foam at the shoulder and 5mm EVA foam at sternum and chest.

Many design considerations go into padded clothing in Rugby Union to affect performance as well as abide by regulations. Flexibility and breathability must be balanced with impact attenuation performance, the location and thickness of the padding must also be considered. As a result of this, significant differences are seen between shoulder padding designs and their performance application as a form of PPE.

2.3 Anatomy

2.3.1 The Shoulder Region

Padded clothing's main area of coverage is the shoulder region. It is therefore important to understand the anatomy of this region. Figure 2.3 details the main structural components of the shoulder area. This illustrates the complexity of the shoulder region.

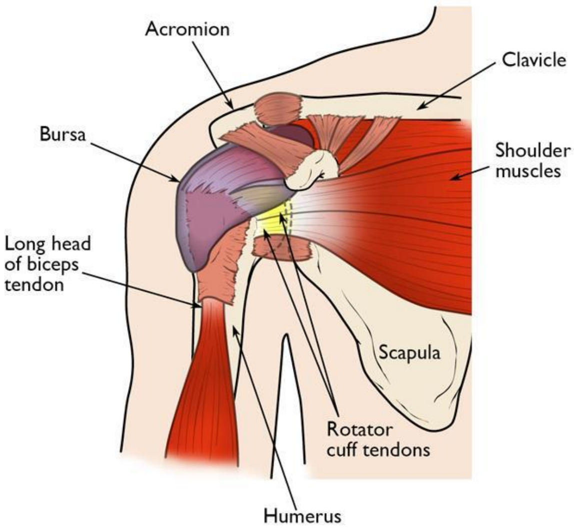


Figure 2.3 – Anatomy of the shoulder region [72].

2.3.2 Skeletal Components

The Shoulder region is made up of three main bones, the Humerus, Scapula, and Clavicle [73]. Bones are a key structural component of the body and are integral in support and movement. The humorous is the largest and longest bone of the upper extremity. It connects the scapula to the bones of the lower arm. The scapula is a triangular bone lying posterolateral to the thorax. It serves as a site of muscle attachment. This protection means fractures only occur through indirect trauma. The clavicle serves as a bony structure connecting the sternum to the shoulder girdle. The clavicle has a double bend along its axis with the middle third being the thinnest portion. This area is also weak mechanically suggesting a high predominance in fractures [74].

There is a relationship between stature and bone length, with many studies defining predictive equations. These equations may be useful in the design of human surrogates, these are presented for the Humerus, Scapula and Clavicle in equations 2.1, 2.2, 2.3 and 2.4. These studies used both male and female particpants from a range of ages and statures.

$$y = \frac{x - 59.9}{3.41} \qquad 2.1\ [75]$$

where y = humerus length (cm) and x = stature (cm)

$$y = 4.247x + 93.74 \qquad 2.2\ [76]$$

where y = longitudinal scapula length (cm) and x = stature (cm)

$$y = 4.68x + 118.5 \qquad 2.3\ [76]$$

where y = transverse scapula length (cm) and x = stature (cm)

$$y = 0.07x + 4.72 \qquad 2.4\ [77]$$

where y = clavicle length (cm) and x = stature (cm)

2.3.3 Muscular Tissue

Skeletal muscle makes up a large amount of the soft tissue in the human limb mainly serving movement and protection. Skeletal muscles constitute around 40 % of the body's mass [78], therefore injury can become regular.

The shoulder complex is made up of superficial and deep muscles (Figure 2.4). A large number of muscles in the shoulder complex mean it can perform a large range of complex movements. The largest muscle in the shoulder is the Deltoid making up 20% of the shoulders musculature [79], it aids in arm abduction, medial rotation of the humorous as well as flexion and extension at the shoulder. It plays a large role in injury prevention as it averts joint dislocation at the Glenohumeral and Acromioclavicular joint [80]. The muscles are attached to the bone via connective tendons to achieve movement. The large number and high volume of superficial muscles are bound to affect the shoulder complex's mechanical response to impact.

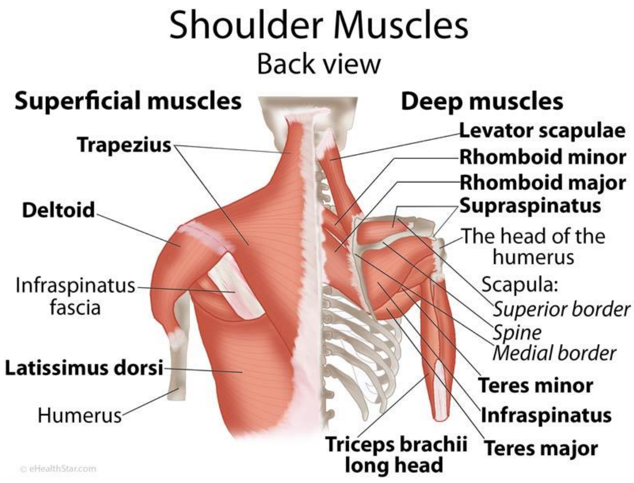


Figure 2.4 – Posterior view of shoulder complex muscles (image from www.anatomynote.com).

2.3.4 Joints

The shoulder complex is made up of four joints (Figure 2.5). The Glenohumeral (GH) joint is a ball and socket joint that articulates between the Glenoid Fossa of the Scapula and the head of the Humerus. The joint is supported by the Glenohumeral, Coracohumeral and transverse Humeral ligaments. This joint is considered the most mobile but least stable in the body making it susceptible to dislocations [81]. The Acromioclavicular (AC) joint is a gliding style synovial joint that articulates between the lateral end of the Clavicle and the Acromion of the Scapula. Its ligaments consist of the Acromioclavicular, Conoid and Trapezoid. The AC joint's main function is to provide the ability to raise the arm above the head [82]. The Scapulothoracic (ST) joint articulates the Scapula with the Thorax. It does not display usual joint characteristics (union by fibrous, cartilaginous, or synovial tissues). The ST joint allows increased shoulder movement beyond the 120° offered solely by the GH joint [73]. The Sternoclavicular (SC) joint articulates the medial aspect of the clavicle and the manubrium of the sternum. Its ligaments include the Sternoclavicular ligament, Costoclavicular ligament, and Interclavicular ligament. This makes the joint very strong meaning fracture of the Clavicle will often happen before SC joint dislocation [83].

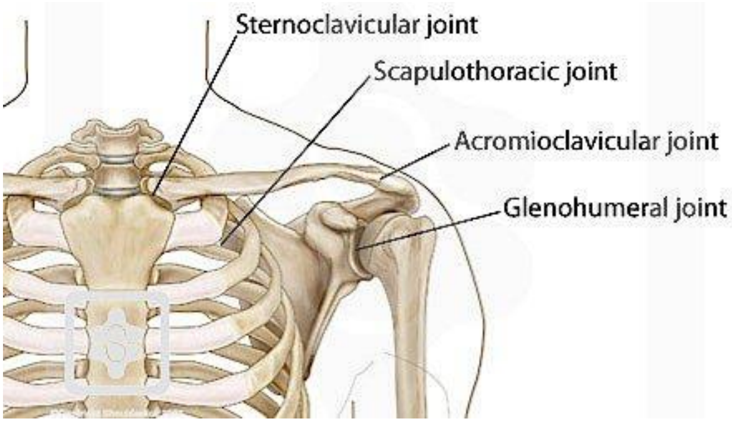


Figure 2.5 – Joints of the shoulder complex (image from www.shoulderdoc.co.uk).

2.3.5 Skin

The skin is the body's outermost layer weighing 3-5 kg in total mass [84]. The skin varies in thickness according to function and the area of the body. Generally, skin is 1-2 mm thick. However, on the eyelids skin is 0.5 mm and up to 4 mm on the soles of feet. The skin consists of two main layers: the Epidermis, and the Dermis. The Epidermis is the external layer that comes into interaction with external surfaces. The Dermis lies under the Epidermis providing physical support and nutrients. The Dermis consists of a network of collagen with interspersed elastic fibers, and lymphatic elements. The structural response is therefore largely determined by the Dermal layer [85]. The skin provides a protective barrier from trauma and therefore its varying thickness, as well as stiffness, will affect its mechanical response to load.

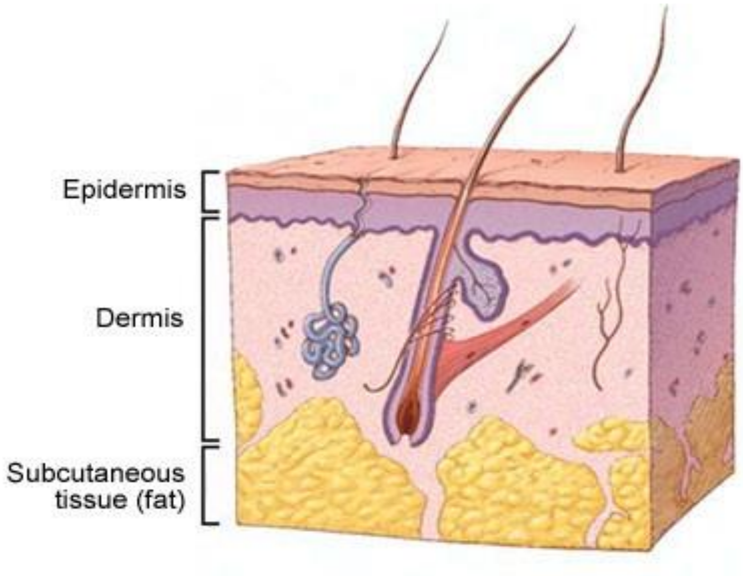


Figure 2.6 – The structure of human skin (image from www.matoclinic.org).

2.3.6 Subcutaneous Adipose

Subcutaneous Adipose tissue (fat) lies between the Dermis and Fascia of muscles or the Aponeuroses. It is strongly joined to the Dermis. Its main role is to store energy in the form of fat. However, it plays an important role as a mechanical load absorber protecting the body from local stresses [86]. Adipose tissues mass is comprised of 60 – 80 % lipid (triglycerides) 5 – 30 % water and 2 – 3 % proteins. The triglyceride is made up of fat cells (adipocytes), these align in ill defines lobules of diameter 1 mm. These lobules are loosely made up of two distinct regions, a basement membrane and an outer sheath of collagen fibers [87]. Adipose tissue has a significantly smaller volume within the shoulder and upper arm when compared to the thigh for example. A study found the upper arm to have 0.29 cm^3/kg of subcutaneous fat less than that of the thigh [88]. Although this is significantly less and due to adipose tissue's shock-absorbing function, it is expected it will affect the shoulder region's response to impact.

2.4 Injury Modes

2.4.1 Overview

To assess the risk of injury and the effectiveness of protection in Rugby Union it is important to understand the commonly occurring injuries that occur to the human body. Many injuries in Rugby Union are caused by external forces like a direct blow from an opponent's shoulder. Contusions, Abrasions, Lacerations, and Fractures are thus classed as acute direct injuries. A Laceration is a type of Cut caused by blunt force trauma to human soft tissue.

2.4.2 Lacerations

In sport, Laceration injuries are very rare. A meta-analysis found Lacerations to account for 5.1 % of all match injuries in Rugby Union. It was also found that Lacerations make up 2.4 injuries per 1000 match exposure hours [89]. It is however important to challenge the injury definition within this study. It is suggested that a time-loss definition is used. Therefore, a player could be treated during match play and an underestimation of Laceration injury prevalence be seen.

When human skin is relaxed, the Collagen and Elastin fibers are unordered. When a load is applied to the skin, it responds by dissipating the energy via its viscous component. This can be explained by Young's Modulus or the stress-strain curve. The stress initially causes the Elastin fibers to carry the load and the Collagen remains unordered. The Collagen gradually aligns in the direction of the load when this increases, eventually causing the fibers to fail. Collagen fibers aligned in the direction of an applied force have been shown to fail at a strain of 5-6 % and pressures in the range of 147-343 MPa depending on body location [90-92].

Lacerations can occur when the tissue is exposed to a crushing force or a sharp object, this leads to local destruction of the tissue structure or even volumetric loss of muscle tissue [93, 94]. Studies have been conducted to establish the mechanical impact forces required to cause Lacerations. Sharkey et al. Sharkey et al. [95] used porcine head cadavers as impact models finding a Laceration to occur at 5259 N when stamping of a trainer. Overall the minimum force associated with the formation of a Laceration was seen to be 4000 N.

However, more representative of a sporting situation, a hammer, and broomstick produced the most Lacerations. This equipment could be perceived to represent a rugby study for example. This also implies objects with a smaller diameter and localised impact pressure is a factor in Laceration injury. The hammer used within this test was of 2 cm diameter potentially representative of a rugby stud. Data suggests porcine skin is alike to human skin although not being completely representative and therefore reducing the ecological validity of the study.

Laceration injuries appear to present a longer recovery period than that of other frequent Rugby Union injuries like Contusions [96]. The use of PPE to prevent Lacerations in Rugby Union, therefore, becomes important in injury prevention, performance enhancement, and financial gain, especially in the professional side of the game.

2.4.3 Abrasions

An Abrasion is the destruction of skin which usually involves the superficial layers of the epidermis only. They are caused by a blow or a fall. Movement and pressure on the surface of the skin are essential [97]. There are four different types of Abrasion highlighted below:

1. Scratches: Caused by a sharp object passing across the skin.
2. Grazes: Movement between the skin and a rough surface – Raking in rugby.
3. Pressure Abrasions: Crushing of superficial layers of skin, can be a bruise surrounding area – Stamping in rugby.
4. Impact Abrasions: Impact with a rough object – Stamping in rugby.

The actual injury mechanics to cause an Abrasion is relatively un-explored. This is possibly due to the ethical issues surrounding testing on humans. Mao et al. [98] developed a dummy skin to simulate skin Abrasion trauma. However, no conclusions on the mechanics to cause an Abrasion was made. There is clearly a need for further research in this field.

2.4.4 Contusions

Muscle Contusions are considered the most common muscle injury in regard to sport [99]. They are caused when the tissue is exposed to a fast and strong compressive force typically from a blunt non-penetrating object. This leads to muscular damage close to the bone. Local Contusions will cause damage to blood vessels causing blood to leak into extracellular spaces. Contusions may then lead to intramuscular bleeding and Haematoma formation [100]. It is suggested that when contracted the muscle is more prone to damage to superficial tissues but when relaxed structural damage and consequently a Haematoma will occur nearer the bone [101].

In order to study the mechanisms causing Contusion injuries experimental animal models and far rarer because of ethical reasons human models have been developed to replicate the process of injury. Desmoulin and Anderson [102] replicated a Contusion injury on a human participant suggesting pressures of up to 4.52 MPa could be experienced with no bruise formation. Far greater than previous studies on animals suggesting the onset of Contusions at 1 MPa [103]. The issue with this is that there are no control measures put in place to evaluate these impacts. Contusion injuries can be influenced by the site of impact, the activation status of muscles involved, the age of the participant, and the presence of fatigue [101]. It is therefore important to consider all these factors when modelling sports impacts.

2.5 Organic Tissue Properties

2.5.1 Overview

The mechanical properties of organic tissues throughout the body have been extensively examined and tested on human and animal samples. The human shoulder region is largely made up of skeletal components, muscular tissue, skin and subcutaneous adipose tissue. These will affect the mechanical response this body region has to load. The mechanical behaviour of these components has been identified.

Ethical considerations, as well as practicality, make it very tough to obtain data regarding mechanical measurements of organic tissue in vivo. Low impact tests designed not to cause human harm have been conducted to establish the mechanical reaction of the thigh to impact with a striker [104]. The issue with this design is that the mechanical properties of whole-body regions or multiple tissues are measured, rather than single body tissues. Other in vivo studies have used surgical scrap consisting of healthy muscles from living human subjects [105]. More practical and intrusive in vitro methods are required to extensively establish organic tissue properties. Testing is done on post-mortem human subjects (PMHS), however, the degradation of the tissue, as well as the difficulty in obtaining samples, prompts a shortness in research [106].

A larger amount of tissue characterisation research has focused on the testing of animal specimens. Porcine tissue has been used as a human tissue substitute. Research suggests porcine tissue displays comparable morphological and mechanical properties to that of human tissue [107]. However, contradicting studies suggest that human muscle tissue displays stiffer properties than that of porcine tissue [108].

As well as the origin of organic tissue specimens, it is important to consider other factors that may affect the mechanical properties the specimen portrays. These factors must be deliberated when carrying out and conducting research:

Site of Specimen

- The mechanical properties of the specimen will vary in regards to the location it was taken. Smalls et al. [109] found skin at the calf exhibits greater stiffness when compared with skin taken from the thigh and shoulder. It was also noted that the location of skin will affect thickness. Similar mechanical differences in skin location have also been found [110]. Balaraman et al. [108] also found a difference in stiffness of muscles at the Soleus and Gastrocnemius.

Age of Specimen

- The age of the specimen can impact its mechanical properties greatly. A study investigating the biomechanical properties of skin found a negative relationship between skin age and elasticity [109]. The age of human bone has also been found to negatively influence mechanical properties like tensile strength [111]. It is therefore important to consider the age of the specimen when applying research in practice.

Hydration Levels of Specimen

- Literature suggests the hydration levels of skin will affect its mechanical properties [112]. However, Hendriks et al., [113] found hydration levels of skin to not affect mechanical response and is all subject dependent.

Temperature at Testing

- The temperature at which the specimens are tested can affect mechanical response. Increased temperature can see a reduction in the Young's Modulus of living tissue [114]. This is a factor that must be considered and controlled in research.

2.5.2 Skin

Skin must be flexible enough to facilitate movement as well as have the ability to return to its original state. The skin can be described as viscoelastic, anisotropic, nonlinear, and non-homogenous [115]. Its properties are mainly based on the concentration and alignment of the collagen fibers. The collagen fibers become very stiff under tension however when under compression they are unable to carry a large load [85].

Collagen and elastin fibers located in the underlying Dermis of the skin, therefore, make up the overall mechanical characteristics of the skin. The mechanical response of the skin under tension has been divided into three parts as seen in Figure 2.7 [116]. In phase 1, fibers are undulated and multidirectional. The skin is largely compliant, and deformation is seen. Through phases 2 and 3, fibers become aligned to eventually becoming fully aligned and straight in phase 3. In turn, the skin becomes stiffer with its behaviour becoming linear.

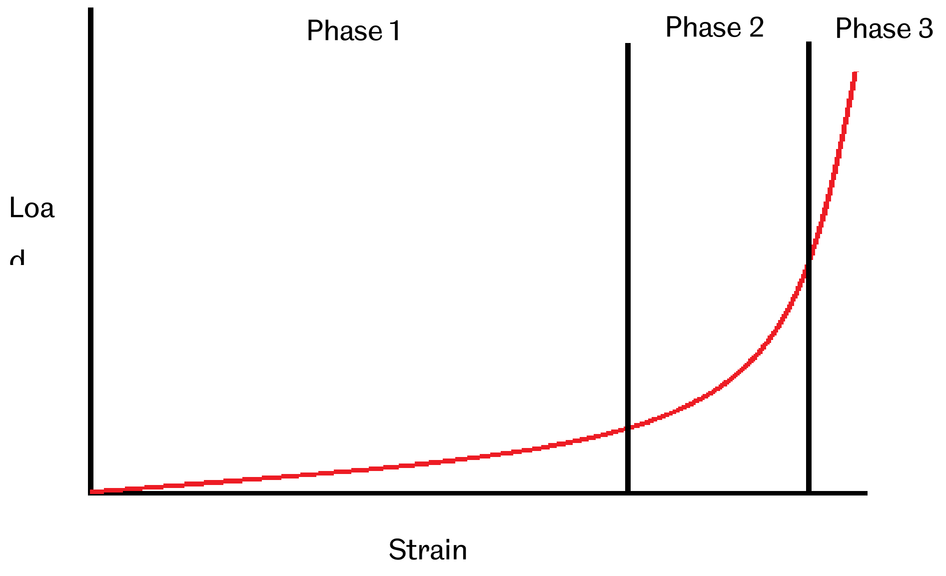


Figure 2.7 – Typical load-strain (tensile) response shown by human skin.

Skin is commonly tested in tension [117]. Some studies have characterised the mechanical properties of the skin in quasi-static loading conditions, other studies have characterised the properties of skin in dynamic loading conditions, this is usually done using a Split Hopkinson Pressure Bar (SHPB) [118, 119]. The results obtained from this research are summarised in Table 2.8. The results vary greatly, this can be put down to differences in the specimen used as well as different loading rates. Younger skin is more elastic than that of older skin [120, 121], this is important to consider in the current research.

Table 2.8 – Summary of skin characterisation research.

Load type	Source	Description	Young's modulus (MPa)	Ultimate tensile strength (MPa)
Dymanic	[119]	In vitro tests on back samples from 77 - 85 year old PMHS.	56.8-141.11	17.9-36.5
Dymanic	[122]	Ex vivo tests on abdominal skin from 43 ± 4 year old females.	2.6 ± 0.6	-
Quasi-static	[110]	In vitro tensile testing of excised human skin.	83.3±34.9	21.6±8.4
Dymanic	[123]	In vitro tests on forehead and arm skin from 85 year old PMHS.	19.5-87.1	5.7-12.6
Quasi-static	[124]	In vitro tests on back and abdomen of porcine skin.	31-53	7-30
Dymanic	[118]	In vitro tests on back of porcine skin.	-	0.1-0.8

Based on impact velocities of 4.3 $m.s^{-1}$ in Rugby Union stud impacts [67] the loading conditions used by Ní Annaidh et al., [110] of 3 $m.s^{-1}$ potentially represent that of a rugby relevant strain rate.

Ballistic research has concentrated on the penetration characteristics of skin. Bir et al. [125] developed a skin surrogate from gelatine. It was concluded that an energy of 23.88 J/cm^2 would entail a 50 % risk of penetration. Other research has used PMHS as a method of estimating the impact energy it takes to penetrate human skin. Bir et al. [126] found a 50 % risk of penetration to occur at impact energies of 23.99 J/cm^2 to 52.74 J/cm^2 depending on body location. These impact energies are similar to total impact energies found in a stamping movement in Rugby Union of 45.52 J/cm^2 when using a stud area of 0.8 cm^2. [67].

2.5.3 Skeletal Muscle

Skeletal muscle must have the ability to facilitate movement in the human body, is it, therefore, viscoelastic, non- linear, anisotropic and heterogeneous. The muscle shows stiffer behaviours when tested perpendicular to the direction of the muscle fibers than parallel [127]. Muscle exhibits greater stiffness when contracted versus being relaxed. Many studies have investigated the mechanical properties of muscle on animals and PMHS.

Song et al. [128] investigated the compressive response of porcine muscle at quasi-static and dynamic loading rates both along and perpendicular to the muscle fiber direction. It was found that the compressive stress-strain response was highly sensitive to the strain rate. Chawla et al. [105] conducted a study on medical scrap from living humans. Engineering stress-strain was graphed, the compressive stress-strain response was also found to be highly sensitive to the strain rate. Bulk modulus (K) was also influenced by the strain rate. At 136 s^{-1} it was found to be 72,680 Pa and at 262 s^{-1} it was found to be 298,600 Pa. Samples were taken from both the shoulder and the thigh region, however mechanical differences between each region were not reported. Balaraman et al. [108] investigated the dynamic response of human Soleus and Gastrocnemius muscle under compression. Specimens were taken from a 43-year-old male and frozen for two weeks before being thawed at 20°C prior to testing. It was found that human muscle was stiffer than of the porcine response reported by Song et al. [128] and bovine response reported by Van Sligtenhorst et al. [129]. However, the increased tonicity found in the muscle tissue of living humans questions the validity of these studies, a different mechanical response may occur in a sporting situation where an athlete's muscle is contracted and braced for an impact.

Research has also quantified the density, ultimate tensile strength, and Young's Modulus of organic muscle tissue. Ward and Lieber [130] measured the density of human cadaver specimens. Mean density was found to be 1.06 g/cm^3 with hydration levels not significantly affecting density.

Yamada and Evans [131] reported a mean ultimate tensile strength of 107 Pa. Data was obtained from PMHS specimens taken from the Rectus Abdominous muscle. The age of the PMHS ranged from 20-39 which is more representative of a sporting population. Friden and Lieber [132] investigated the tensile strength and Young's Modulus on human upper extremity muscle biopsies. Mean ultimate tensile strength was 9000 Pa and mean Young's Modulus was 28.25 kPa. Ultimate tensile strength was found to be greater in muscle biopsy specimens rather than PMHS specimens. This maybe because specimens were taken from different body locations as well as the degradation considerations that are invoked with PMHS specimens.

Many studies reporting organic muscle properties vary in results. Many studies use different strain rates which makes a comparison between studies problematic. Strain rate ranges that are relevant to rugby impacts must be considered.

2.5.4 Subcutaneous Adipose Tissue

Subcutaneous Adipose tissue is a soft connective tissue located directly under the Dermis of the skin. Adipose provides thermal insulation, energy storage, and shock protection while having the ability to allow for the free movement of underlying muscle groups. It, therefore, possesses a low stiffness and demonstrates viscoelastic, non-linear and heterogeneous mechanical properties [133, 134]. Adipose tissue is generally characterised using human specimens from fat pads on the heel or hand as well as breast tissue and porcine tissue [86, 133, 135, 136].

The density of human Subcutaneous Adipose tissue has been reported to be 0.93 g/cm^3 [137]. Many studies have investigated the mechanical response of adipose tissue. Samani and Plewes [135] conducted in vitro tests on human breast adipose tissue. The Young's Modulus was found to be 3.6 kPa. Geerligs et al. [138] investigated the mechanical response of porcine adipose tissue, the shear modulus was found to be 7.5 kPa at 10 rad/s and 37°C. Erdemir et al. [139] carried out in vivo indentation tests on the human heel pad. The compressive modulus was found to be 49.4 kPa. This is far greater than that of the previous studies mentioned. It is suggested this is because they display morphological differences because of the high-stress levels involved in their anatomical function [140].

As with other organics tissues, it is understood that adipose exhibits a nonlinear response to load and is significantly rate-dependent. Alkhouli et al. [141] examined the mechanical properties of human subcutaneous adipose over varying strain rates. At 30 % strain, the Youngs Modulus was found to be 1.6 ± 0.8 kPa. From 30 % to failure Young's Modulus increased to 11.7 ± 6.4 kPa. Comley and Fleck [133] reported a similar response. Young's Modulus increased from 2 kPa at a strain rate of 10 s^{-1} to 4 MPa at a strain rate of 3000 s^{-1} on porcine adipose tissue specimens.

Adipose tissue represents a varied mechanical response dependent on the strain rate in which it is deformed. It is, therefore, appropriate to use studies that represent a wide range of strain rates especially applying these to rugby impact situations when fabricating a human impact surrogate.

2.5.5 Bone

The mechanical properties of bone are important in injury biomechanics research as well as the development of surrogates. Their failure loads and impact response must be quantified. The bone is a composite material made up of Collagen and Hydroxyapatite. The Young's Modulus is intermediate of between that of Hydroxyapatite and Collagen. Therefore, the bone's mechanical properties are dependent on the concentrations of Hydroxyapatite and Collagen as well as the geometric shape and bonds between fibers and matrix. The bones strength is higher than both Hydroxyapatite and Collagen because the softer component (Collagen) prevents the brittle component (Hydroxyapatite) from brittle cracking, while the stiff component prevents the soft component from yielding [114].

Bone is a hard material and stress-strain relationships can be made similar to many engineering materials. Cortical bone and Trabecular bone make up their two structure types. Cortical bone is present in the shell and central section of the long bone while Trabecular bone makes up the 3D lattice within the inner surface of the Cortical bone [114].

The density of organic Cortical and Trabecular bone varies. Cortical bone values have been found to be 1.47 – 2.12 g/cm^3 [142] and 1.7 – 2.0 g/cm^3 [143]. Trabecular bone values have been found to be 0.18 – 0.95 g/cm^3 [144] and 0.13 – 0.69 g/cm^3 [145]. Table 2.9 summarises research that has studied the mechanical properties of bone.

Table 2.9 – Summary of bone characterisation research.

Source	Description	Youngs modulus (GPa)	Ultimate tensile strength (MPa)
[131]	In vitro quasi-static test on human femur of PMHS aged 20-39 years.	17.6	124 ± 1.1
[131]	In vitro quasi-static test on human humerus of PMHS aged 20-39 years.	17.5	125 ± 0.8
[131]	In vitro quasi-static test on porcine humerus bone.	14.6	88 ± 7.3
[146]	In vitro test on 20- 29 aged PMHS specimens from human femur.	17 ± 2.2	140 ± 10
[146]	In vitro test on 30- 39 aged PMHS specimens from human femur.	17.6 ± 0.3	136 ± 3.5
[146]	In vitro tests on PMHS specimens from human tibia.	19.9 ± 2.4	147 ± 9.2

As with muscle characterisation research, stiffness and failure values of bone tissue have been shown to be rate-dependent. This is shown in a study by McElhaney [147] who found Young's Modulus to increase from 15.2 GPa to 40.7 GPa between strain rates of 0.001 – 1500 s^{-1} in a PMHS 24-year-old femur specimen. The wide range of strain rates in this study represents data that is suitable to rugby impacts and must be considered in the current research. The existence of data specific to strain rates that are representative of organic tissue behaviour in sports impacts with Rugby Union, in particular, would produce datasets beneficial to the development of a human shoulder surrogate.

2.6 Human Tissue Surrogates

2.6.1 Overview

To understand injury biomechanics, it is important to develop accurate human tissue surrogates. From this, an increased understanding of the mechanisms of injury, characterisation of human response to these loading conditions, and the effectiveness of protective clothing can be assessed. These surrogates can be divided into organic and artificial, each varying in biofidelity, durability, suitability, cost, and ethical application, bringing specific strengths and weaknesses [11, 148]. Traditionally, human participants, PMHS and animal surrogates have been used in injury biomechanics research. However, technological advancements have seen the development of synthetic surrogates and computational models, eliminating factors such as ethical constrictions because of the use of biological tissue as well as repeatability issues due to the lack of durability of the surrogate.

Several industries have utilised human tissue surrogates to better understand injury biomechanics. Namely ballistics, automotive research, and in the current research, sports PPE have developed artificial surrogates that are validated by organic surrogate research. From this, notable applications can be seen like improved injury prevention [11].

2.6.2 Organic Surrogates

Organic surrogates refer to any biological tissue that aims to replicate the response of living humans, these can be living or postmortem. Organic surrogates are often used to determine the mechanical properties of human tissue as well as their response to impact. Organic surrogates are regularly used to validate and develop artificial surrogates.

Human Participants

Although not a simulant, human volunteers can provide the most accurate representation of the mechanical response of human tissue. The primary advantage of this is that research can carry out tests in vivo without having to make any assumptions regarding the internal structure, tissue composition, and physiology. However, severe ethical implications are linked with this approach. Research using human participants must be performed at non-injurious levels as well as exert no pain to the subject in accordance with the Nuremburg Code of 1947 and the Helsinki Declaration of 1964. Human testing must also go through strict ethical guidelines making research time-consuming, costly, and most importantly loading mechanisms that cause tissue failure cannot be applied creating an issue with predicting human response to high-intensity loading from low-intensity loading.

In sport, human volunteers have been used to predict responses to impacts. Studies have been set up in a laboratory environment to establish the kinematics of human movements in response to impacts [34, 149]. These studies lack ecological validity as laboratory environments can vary from an 'in-play' situation.

Other studies use human participants to model human tissue's response to impacts. Tsui & Pain [150] used an impactor to explore the effects of muscle tension on biomechanical response to impacts. Alternatively, Hrysomallis [104] completed a study using human participants to validate an artificial surrogate by dropping an impactor on the participant's thigh. Although using human participants will give an exact representation of mechanical response to impact, an injury-causing load could not be applied.

The ability to test a population at injurious loads would pose the ultimate solution in which to investigate sports PPE. However ethical legislation means this is not possible. Therefore, although useful in some scenarios the use of human participants becomes practically and ecologically invalid.

Post-Mortem Human Surrogates

PMHS, also known as cadavers possess that of human tissue structures that can be found in in vivo human subjects. The mechanical properties of organic tissues throughout the body have been extensively examined on PMHS, however, in regards to modelling sports impacts PMHS are rarely used [151]. Hrysomallis [104] used cadavers to develop an artificial surrogate thigh. Heald & Pass [152] used head cadavers to explore injury mechanics to the head from a baseball.

PMHS crucial strength is that they possess the exact anatomical structures found in living humans [153], with no current artificial surrogates having the ability to replicate this [151]. This makes PMHS valuable when assessing human tissue damage.

However, PMHS come with many limitations, The PMHS lack of tonicity leads to a lack of physiological response to impact. This greatly decreases their biofidelity [148]. Because cadavers are in vitro biological structures, there is an amount of decomposition before testing which can change their mechanical properties [154].

The age of PMHS specimen is also questioned. Cadavers are generally sourced from an elderly population. In vivo testing found skin elasticity to decrease with age [155]. The age of human bone has also been found to negatively influence mechanical properties like tensile strength [111]. It is therefore unlikely that the characteristics displayed by cadavers will be representative of an athletic population, especially at an elite level.

Due to ethics, the availability of PMHS specimens becomes limited [153]. Because of this, specimens tend to be impacted multiple times, this can lead to PMHS specimens becoming damaged consequently altering their mechanical response between trails [11]. As well as this, cadavers are regarded as a level 2 biohazard. Trained personal and specialist equipment are therefore required to handle and experiment on PMHS specimens, thus increasing the cost of any experiment [148].

Although cadavers pose a suitable surrogate for assessing the mechanical properties of human tissue, they are not suitable for assessing the standards of PPE due to the social, ethical and cost issues associated with PMHS specimens.

Animal Surrogates

Animals have been used as a surrogate in injury research extensively. Their primary advantage is the ability to test at pain-inducing loads which is not possible on living humans because of ethical reasons. Although the testing on living animals is possible with primates displaying the most comparable anatomical and mechanical properties, the use of deceased commercially farmed animal specimens is more common because of social acceptance and ethical concerns. However, animal cadaver specimens have the same issue that human cadaver specimens display. This is mainly because of the difference in physiological response between live and deceased specimens [148].

The testing of anaesthetized live animals also has many issues. There are very strict ethical considerations when testing on live animals which increase cost. Coupled with this, issues arise from public groups against animal testing. Although testing on live animals would be considered an accurate surrogate, clear issues mean it would be unfeasible to use live animals to model biomechanical injury [148].

Deceased porcine skin is often used as a skin simulant because of its similarity with human skin as well as low cost and availability [156]. It has been used previously as a skin surrogate in sports equipment testing as well as to validate artificial surrogates [157]. However, the use of deceased biological tissue can be unhygienic, which is a complication when considering specimens for sports impact surrogates. Specimens are also susceptible to rapid degradation, variation, difficulties with regards to the details of the specimen’s origin as well as their mechanical properties generally changing after one impact reducing their repeatability. [158].

Animal surrogates do not display the exterior geometries of humans with primates being an exception. However, their similarity to the mechanical properties of some human tissue makes them suitable in sports injury evaluation.

2.6.3 Artificial Surrogates

Artificial surrogates refer to any inorganic models of human tissue. They are developed using the data obtained from organic surrogates. This data is then used to create a surrogate that represents the mechanics of living humans. Generally, they can be split into two types: synthetic surrogates and computational models.

Synthetic Surrogates

Synthetic surrogates present a feasible option for use in injury biomechanics research when organic surrogates cannot be used. They pose a solution to issues with organic surrogates like ethical considerations, cost, tissue degradation, and reproducibility. More importantly, they allow a researcher to study impact response without physically harming a human or animal [11]. Synthetic surrogates cannot currently replicate injury to soft tissue because of the structure's complexity. However, they can accurately model the mechanical response seen at impact providing a controlled and repeatable assessment of the injury phenomena and protective equipment.

Although synthetic surrogates lack biofidelity, their reproducibility and sensitivity are notable benefits. Sensitivity refers to the ability of the surrogate to show a different response when the loading criteria are changed. Reproducibility refers to the ability of the surrogate to show the same response when the loading criteria are repeated. This allows the surrogate to produce a standardised response while also allowing a researcher to compare between trails [124].

Generally, synthetic surrogates are validated by comparing their mechanical response to organic surrogates [153]. Force-time curves and stress-strain curves attained from organic surrogates are commonly used as validation data [159, 160]. When using synthetic surrogates for impact research, durable and frangible surrogates are used, each has its benefits and limitations with regards to biomechanical injury research.

Within sports PPE research durable surrogates are commonly used. They offer a multi-use surrogate that presents repeatability. Mechanical response parameters like force and accelerations can be measured during impacts and can be linked to PPE performance. Injury likelihoods can also be made [160]. Crandall et al. [148] split their biofidelity into internal and external. Internally the surrogate attempts to replicate the deformations and accelerations seen in a human tissue impact, externally the surrogate attempts to replicate the interactions seen with the environment at impact.

Within sports PPE there has been no research where a durable surrogate has been used to model the upper body or shoulder. However, there is limited research when considering the lower body. Hrysomallis [104] manufactured a surrogate thigh model to test cricket thigh protectors. Data from human volunteers and PMHS were used to validate the model. Hrysomallis used the silicone Silastic™ 3483 to model human tissue. When compared to a criterion, peak deceleration values were very similar, however, this surrogate only used one material to represent all of the soft tissues in the thigh. This could cause issues with impact response as the skin and muscle will show different response mechanisms.

More recent research by Payne et al. [151] also developed a durable thigh surrogate. Three layers of Polydimethylsiloxane (PDMS) silicones were used to model the muscle, adipose and skin tissues in the thigh. The PDMS simulant for each soft tissue layer was developed using PMHS and animal data. Although this research represents a thigh surrogate that can model impact response accurately, various limitations must be considered. The effects of muscle contraction have not been considered; different skin layers were not modelled as well as the use of in vitro methods for characterisation of tissue simulants. These should all be considered in the current research.

Synthetic skin simulants have been developed to mimic the mechanical properties of skin, with various materials (Liquid suspensions, Gelatinous substances, Textiles) being used depending on the goal of the research [161]. Skin simulants have been developed for skin grafting, skin tribology and injury research. Synthetic chamois leather has been used as it simulates the frictional and mechanical behaviour of skin. Chamois leather has been successfully used in ballistic [125] as well as sports injury research [157]. Silicone rubbers have also been used within biomechanical injury research to simulate the skin. Whittle et al. [162] used silicones to study blunt force trauma. A deguform silicone was used to simulate the skin. Although a successful skin and underlying human tissue model was fashioned its veracity can be questioned as it was not validated.

Computational Models

The use of computational models in injury research has amplified in the present time due to technological advances allowing complex sophisticated models of the human body, as well as the ability to simulate a wide range of conditions and accurately predict injuries. Coupled with this, its capacity to model injury mechanics without causing harm to humans and the need for zero physical materials make it a plausible alternative to using cadavers, animals, and synthetic simulants [148]. Computational models have previously been categorised into three categories [163].

1. Lumped-mass models - simple models of masses linked by springs and dashpots representing basic mechanical response.
2. Rigid-body models - more complex than lumped-mass models where rigid bodies are connected by mechanical joints. The model attempts to simulate human body structure, joint mechanics, and mass distribution.
3. Finite element (FE) models - a model mesh is used to interconnect each element of human tissue based on its mechanical properties.

The simplicity of lumped-mass models means they are rarely used in injury biomechanics research, although they have been used to simulate responses to neck impacts in car crash events [164]. Rigid-body models are also rarely used in injury biomechanics research because of their inability to model body deformation and tissue failure. However, their balance between cost and computational power has seen them used in the automotive industry regularly [165].

FE models represent tissue-level response and failure by the use of governing equations based on concepts developed in previous mechanical research [148]. This meaning they are used regularly within injury research despite the required computing power. FE models provide the ability to accurately predict local stresses and strains using set external loading conditions, these loading conditions can then be refined to different injury-causing loading conditions tested at maximal and sub-maximal injurious levels, leading to a greater understanding of human tissue failure under different loads [163]. However, FE models are heavily dependent on the input and simplifications used to model the geometries and material properties [166]. For a model to be accurate, these inputs must be validated against experimental data, currently, there is insufficient organic data to do this. Presently FE models will therefore only approximate human response based on a set of simplifying assumptions [148, 163].

Although a large amount of FE models are used in automotive injury research, there is some research directed at sports injury. Many FE models have been fashioned to simulate head injury. Patton et al. [167] developed an FE model to investigate brain tissue deformations during elite Australian Rules Football, Rugby League, and Rugby Union. The FE model was optimised using PMHS and real kinematic data in which concussions were and were not sustained, however it was validated using data from car crash situations rather than sports impacts meaning experimental conditions could not be matched precisely.

Payne [159] developed an FE model of a human thigh to provide a simplification of actual human impact behaviour from which sensible predictions of human impact response can be estimated and then validate a silicone model of the thigh that aimed to model sports impacts and assess the performance of sports PPE. Using an FE model to predict that a synthetic surrogate behaves similarly to organic tissue can provide a beneficial comparison tool, however, the testing of impact response with real humans would provide the superior method of evaluating the surrogate’s biofidelity.

CHAPTER 3 - SPORTS IMPACT ASSESSMENTS AND WORLD RUGBY REGULATION 12

3.1 Chapter Overview

This chapter outlines the key components for assessing sports impacts. It then critically reviews the current regulation, “Regulation 12 Schedule 1: Specifications Relating to Players Dress, Sections 5-6”, set out by World Rugby™. Using this, a conceptual model has been applied for rugby impacts to the shoulder. This information as well as a survey study entitled Perceptions and Attitudes Towards Shoulder Padding and Shoulder Injury in Rugby was then used to establish the research approach taken to develop an improved Regulation 12.

3.2 Introduction

3.2.1 Components of Sports Impact Assessments

Sports padding impact assessments and testing feature many variables, dependent on various factors, the sport (i.e., Rugby Union) and the type of impact (i.e., a shoulder tackle) being examples of this. However, there is a sequence of key components that are fundamental in all assessments:

1. **Test Parameters:** The kinetics and kinematics of the striker and target surrogate are seen immediately before and after the impact.
2. **The Surrogate:** Commonly named an anvil, this is the target body the protection (padding, helmet, shin guard) will sit on, it will often represent the human body structure the padding is aimed to protect and is usually a fixed component.
3. **The Striker:** This is the striking element the padding will be impacted by. It is a moving component and will represent a key element relevant to the sports impact scenario, for example, a stud.
4. **Assessment Procedures:** The methods of assessment used to measure the performance of the padding, for example, peak impact acceleration will be evaluated and measured using an accelerometer.

3.2.2 Review of Previous Sports Injury Models

Many models have been published that attempt to represent a sports injury framework. These models can be defined as a framework that explains the processes in which sports injury occurs. It is important to review these models to guide the formulation of a rugby impact model and therefore an improved testing protocol to Regulation 12. These models have been categorised into three approaches in recent sports injury research [168]:

1. **Risk accumulation and intensification model:** The injury occurs because of an accumulation of risk factors.
2. **A mechanical phenomena sequence:** The injury occurs because of a sequence of mechanical loading events.
3. **An event sequence entity matrix:** A matrix of important elements that can cause an injury.

a) Risk accumulation and intensification model

Meeuwisse [169] developed the first risk accumulation and intensification model named the 'multifactorial model'. This was further updated into a more dynamic model later [170] (Figure 3.1).

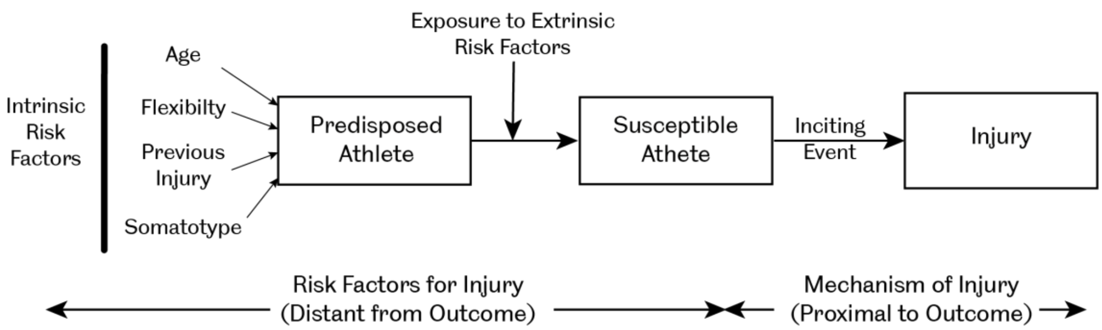


Figure 3.1 – Multifactorial Model (Adapted from Meeuwisse [170]).

The model outlines the principles relating to the assessment of risk factors and causation of sports injury, applying this model will lead to increased success in predicting injuries in sport as well as guiding effective preventative strategies i.e., padding design and regulation. The model has further been adapted, mainly by researchers identifying specific risk factors for specific injuries [171]. Meeuwisse [170] then added a dynamic framework that considered how athletes would adapt to previous impacts as well as recover from previous injuries, hence shifting the risk.

Wismans [172] presented research that highlights a series of mechanical loading events that lead to injury (Figure 3.2). Put simply, it suggests that when a body (human body part) is put under external loading it will deform, this triggers a biomechanical response that can be different between people. Injury preventative measures can be put in place to change this biomechanical response. The body may deform past a recoverable limit and therefore past injury tolerance levels which will, in turn, cause an injury. McIntosh [19] takes this further by outlining that training, coaching and psychological factors can modify injury tolerance levels.

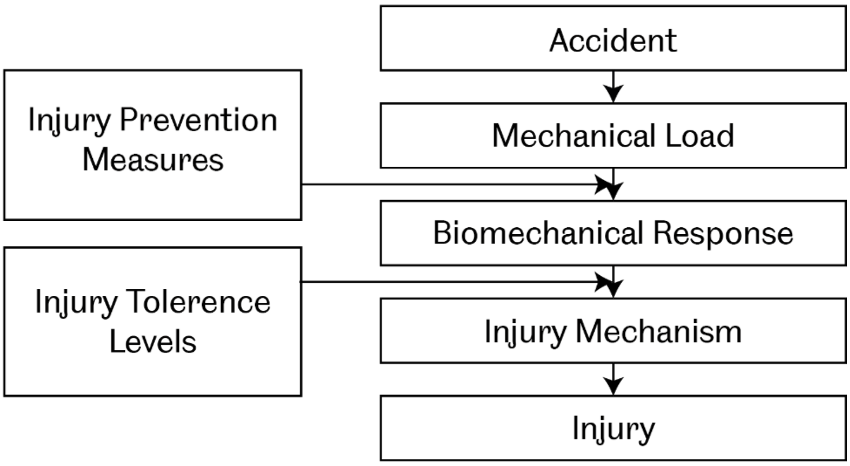


Figure 3.2 – Load Injury Model (Adapted from Wismans [172]).

Haddon [173] first developed a matrix that describes the sequence of events that lead to injury. It consists of a nine-cell matrix that each describe important elements leading to injury, an injury preventative measure could be added to each element. Although initially used in the automotive industry it has been applied in a sporting context. Bahr & Maehlum [174] applied this using the headings; environment, human, equipment and has been used to describe injuries in American football. An example of Haddon's matrix is seen in Figure 3.3, adapted for a rugby setting.

	Host i.e., athlete	**Agent i.e., rules, equipment**	**Environment i.e., sports setting**
Pre – event	Fitness levels	Tackle laws	Medical staff on site
Event	Technique of tackle	PPE	Pitch conditions
Post – event	Reporting of injury	Exposure to repeat trauma	Injury documentation

Figure 3.3 – Haddon Matrix in a rugby setting (adapted from [175]).

Each model aims to describe the events which cause an injury; however, each model is missing key factors which may lead to this. The risk accumulation model does not cover the events before preceding injury, it also does not mention the body's response to impact. The mechanical phenomena sequence model is quite vague, and the Haddon matrix fails to give enough information to distinguish between each entity leading to injury. Each model provides a good basis for better describing an injurious event however their academic nature means they are not applied to specific injurious scenarios. However, many researchers have adapted these models for specific injuries, Bahr & Krosshaug [171] conceptualised ACL injuries for example.

It is important to review previous sports injury models to provide a guideline for subsequent studies and form an impact model that can be used for shoulder impacts in rugby. The following work presented in this chapter aims to:

1. Critically review the current test protocols used in Regulation 12 by replicating test procedures on current padding clothing samples.
2. Develop a shoulder impact model specific to rugby that describes the events and factors leading to a shoulder impact, whether this causes injury or not.
3. Develop an understanding of rugby players' attitudes towards shoulder padding to guide an improved Regulation 12 and add original research to this field.
4. Use the model to set out a framework that aligns the subsequent research to develop an improved Regulation 12 and measure the performance of shoulder padding.

3.3 Critical Review of Regulation 12

Before looking to improve the test standards of shoulder padding as well as assess their performance it is important to review the current standards to date and consider the research questions outlined by World Rugby™ at the start of the PhD.

RQ1. Is the current requirement for padded clothing appropriate for the modern game of rugby, how and why?

RQ2. Is the current requirement for padded clothing appropriate in permitting the use of modern technology, how and why?

This section, therefore, critiques certain elements to the Regulation 12 document as well as details and reviews impact testing completed on commercial padding using the current testing standards.

3.3.1 Critique of the Regulation 12 Document

After an examination of Regulation 12, many sections have been highlighted for review, detailed below:

I.

In Section 5.2.1 - Material Construction, it states "It is the manufacturer's responsibility that all materials used should not be adversely affected by water, dirt, perspiration, toiletries, household soaps and detergents. All materials coming into contact with the wearer's body will not be of the type known to cause skin disorders and shall not cause Abrasion of either the wearer or other players."

Regulation 12 defines that a material should not be affected by certain substances but no standards for this are mentioned against which these properties can be tested. Recommended standards and testing procedures for materials and their response to the mentioned conditions include ISO-15487[1] and ISO-22958[2]. These could be included in the regulation.

II.

In Section 5.2.2 - Padding Materials, it states "Padding materials must be homogeneous (i.e. padding facing towards the wearer must be the same texture, hardness and density as that facing the opponent). Foam padding of sandwich construction is not allowed."

A requirement to test in both directions would allow manufactures to develop non-homogenous designs, while ensuring there is no advantage to the wearer and injury risk to the opposition does not increase. Therefore, permitting the use of more modern materials. The use of sandwich construction is also not permitted, these types of foams are generally lightweight but can be stiff [176]. This could give the wearer an advantage and increase injury risk, the density would also be difficult to measure. If the padding were tested for its stiffness in both directions and passed impact test protocols, sandwich designs could be permitted.

[1] ISO 15487:2018: Textiles -- Method for assessing appearance of apparel and other textile end products after domestic washing and drying

[2][2] ISO 22958: Textiles -- Water resistance -- Rain tests: exposure to a horizontal water spray

III. Section 5.3 - Design

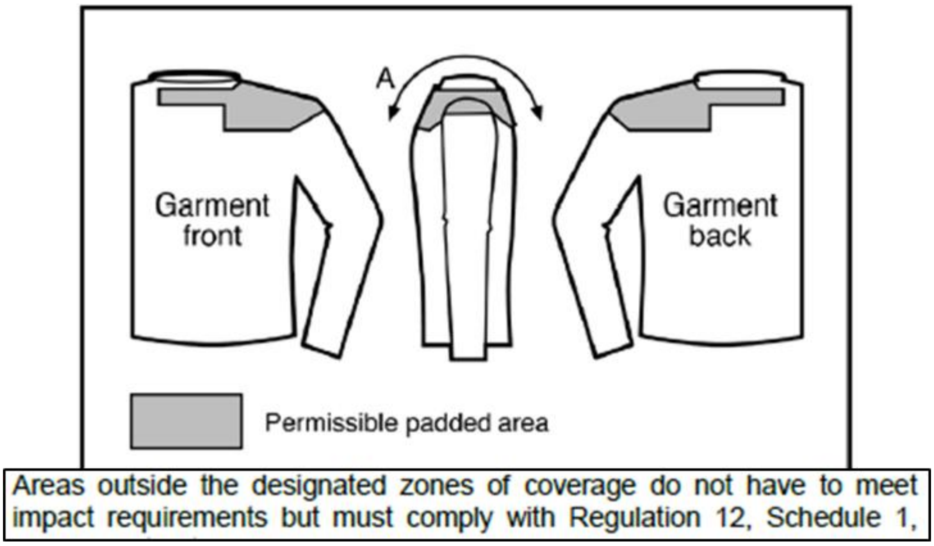


Figure 3.4 - Image from Regulation 12: Areas of Coverage and statement allowing other padded areas to forgo impact attenuation testing.

This section specifies the areas in which padding is permitted on the padded clothing garment, as well as the thickness and density requirements of the padding material. it has two defining restrictions:

The zones of coverage: The document defines the permissible area of coverage while also allowing padding outside this area of coverage but with no impact requirements. However, maximum padding thickness outside the zone of coverage is restricted to 5 mm (Figure 3.4). Padded material can therefore be placed outside this zone with no impact performance regulations and the effect this could have is unknown.

Padding material: Padding density is restricted to 45 + 15 kg/m^3. The addition of +15 kg/m^3 (+33 %) allows padding density to be anywhere between 45 to 60 kg/m^3. Density is also the only physical property controlled by the regulation, using a parameter like stiffness could help to restrict materials that could cause harm to a player.

iv. Section 5.3.3-Sizing

The sizing section defines the nominal chest dimensions (i.e S, M, L) rather than a range for each size, which allow the zone of coverage for padding to be varied. These dimensions do not consider the anthropometric variations that occur with geographical variations. Adapting these sizes based on standards for clothing sizing (such as BS 6185:1982[3] in UK, or JIS L 4004[4]

[3] BS 6185:1982 - Specification for size designation of men's wear
[4] JIS L 4004:2001 Sizing systems for men's garments, Japanese Standards Association

(2001) in Japan) used across the world would allow better sizing and fit, as well as regulation of the zone of coverage for padding.

v.

In Section 5.4.1 - Performance Requirements - Impact Attenuation it states "When tested in accordance with the procedures specified in Section 6.3, the peak acceleration of impacts delivered to test locations shall not be less than 150g."

The impact performance requirements limit the maximum amount of protection the padding can offer while not limiting the minimum amount. The peak acceleration value (150 g) does not have reference to any supporting research. Using a value that is supported by research as well as adding a minimum impact protection requirement would make for a better regulation.

vi. Section 6.2 - Condition of Specimens

Testing must be completed at two temperatures of 20° C and 50° C. these temperatures may not be applicable to the temperature of padding in game play. Understanding these temperatures by simulating in-game environments may give an indication of a more appropriate conditioning temperature of the padding at testing. The maximum time (5 minutes) between removal of padding from a conditioned environment (i.e. an oven) to testing could also be reduced to improve repeatability.

vii.

In Section 6.3.2 – Apparatus it states "The apparatus for the impact attenuation test shall consist of the following (also see Figure 9):

Drop Assembly - a dropping mass shall be attached to a free fall or rail guided drop assembly carriage. The mass shall be 5 kg +/- 0.02 kg. The dropping mass shall have a flat striking face of diameter 130mm +/- 2mm.

Anvil- the anvil shall consist of a horizontal steel cylinder with a diameter of 115mm +/- 2mm and shall not have a resonance frequency liable to affect measurements. The centre of mass of the drop mass shall lie over the centre of the anvil."

The drop mass (striker) and anvil (surrogate) are rigid. This setup is unrepresentative of human tissue strictures and may not represent the shoulder impacts seen in rugby. This is even more unrepresentative when considering injuries like Cuts, Lacerations and Abrasions. The test protocols seek to control the impact attenuative properties of the padded material.

However, using a more biofidelic surrogate and/ or striker would help to improve understanding of the impact performance and protective capabilities of the padding.

viii. Section 6.3.4 - Impacting

The impact testing methodology is explained but the number of impacts is not defined. A clearly defined and detailed methodology should be included to ensure consistency in testing of padding between laboratories/ test houses.

3.3.2 Replication of Test Procedures

Eleven designs of World Rugby™ approved padded clothing samples were obtained from five manufacturers (sampled were anonymised for confidentiality) and intact padding samples (size range: 150 x 120 mm to 280 x 220 mm due to different manufacturer design) were taken from the padded area. A control material (Aortha White, Plastazote®, -LD-60, Algeos), similar to the foam used in shoulder padding due to its thickness (10 mm) and density (LD-60 corresponds to 60 kg.m^{-3}) while also meeting the requirements of Regulation 12 was also used. A 220 x 150 mm sample was cut from the control material to test alongside the padding samples.

The impact test setup can be seen in Figure 3.5 and can be used to replicate the current procedure used in Regulation 12. Further details of this drop rig are explained in §6.2.

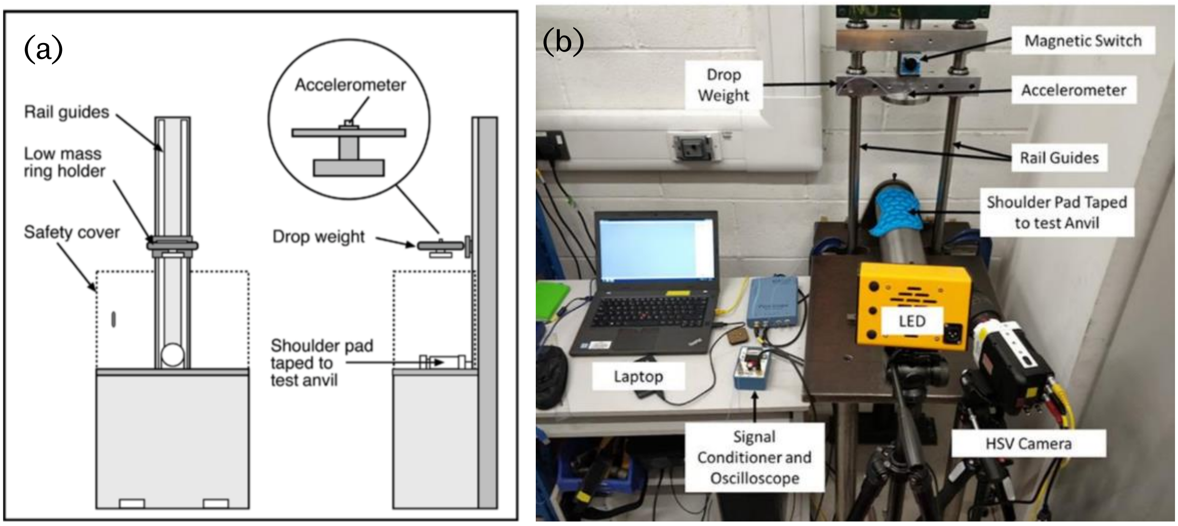


Figure 3.5 – (a) Drop rig schematic described in Regulation 12 (b) Drop rig used for impact testing as per Regulation 12.

Two samples of each shoulder padding were cut from the material that encased them and the thickness was measured at 3 different locations across the padding using a Digital Calliper. One sample was heated in an oven (LHT6/30, Carbolite Gabe Ltd, UK) to 50° for 4 hours, while the other sample was maintained at room temperature (20 ± 2°C) as per Regulation 12. The temperature was measured straight before impacting using an Infrared Thermometer (TestSafe, TS-TM001, Burton-on-Trent, UK).

Each sample was then impacted at an impact energy of 14.7 J (5kg mass from 30cm height) using the impact rig shown in Figure 5.2(b). The striking mass had a flat circular face with a diameter of 130 mm. Each sample was fixed to a horizontal steel cylinder anvil of a diameter of 115 mm. The striking mass was fitted with a single axis accelerometer (352B01-ICP-Accelerometer, PCB Piezotronics) sampled at 200 kHz and connected to an oscilloscope software PicoScope®(Version 6, Pico Technology) via an ICP® sensor signal conditioner (480B21, PCB®), to enable temporal acceleration to be obtained throughout impact. All filtering was completed automatically by the signal conditioner, therefore no manual filtering was needed. The impact was also filmed with a High-Speed Video (HSV) Camera (Phantom Miro 34 R111, Vision Research, USA) with a resolution of 512 x 320 at a sample rate of 10kHz and 99.00µs exposure rate and was synced to the accelerometer using the PicoScope.

Each sample was impacted 3 times with a 1-minute recovery between each impact. This allowed testing to be completed within 5 minutes of removing the sample from the oven as per Regulation 12. Peak acceleration for each impact was calculated and results analysed.

Results

Impact testing as per Regulation 12, on 11 samples plus 1 control showed varied peak impact force attenuation capabilities, as seen in the data provided in Table 3.1. Apart from the control material (sample 12), all materials exceeded the minimum peak acceleration value of 150g. The peak forces of the impacts are all much greater than what is seen in a rugby tackle on the pitch (3400 N, Dynamic Tackle [62]). This is probably due to the rigid nature of the striker and anvil.

Table 3.1 – Thickness and Peak Acceleration results as per Regulation 12.

Sample No	Temperature 20 ± 2°C			Temperature 50 ± 2°C		
	Thickness (Mean ± SD) (mm)	**Peak Acceleration (Mean ± SD) (g)**	**Peak Force* (Mean ± SD) (N)**	**Thickness (Mean ± SD) (mm)**	**Peak Acceleration (Mean ± SD) (g)**	**Peak Force* (Mean ± SD) (N)**
1	9.9 ± 0.1	220 ± 1.2	10787 ± 59	9.9 ± 0.2	206 ± 2.4	10080 ± 119
2	10.9 ± 0.3	221 ± 0.7	10836 ± 34	11.0 ± 0.5	199 ± 1.9	9768 ± 95
3	10.1 ± 0.1	229 ± 1.1	11229 ± 54	9.9 ± 0.1	203 ± 2.9	9942 ± 143
4	9.7 ± 0.2	223 ± 2.9	10934 ± 142	10.0 ± 0.2	203 ± 2.2	9944 ± 110
5	9.8 ± 0.4	232 ± 1.3	11375 ± 64	9.9 ± 0.3	193 ± 4.5	9477 ± 220
6	10.6 ± 0.2	223 ± 1.4	10934 ± 69	10.0 ± 0.7	207 ± 1.3	10149 ± 63
7	8.1 ± 0.2	210 ± 5.2	10297 ± 255	10.8 ± 0.6	195 ± 5.1	9572 ± 248
8	8.1 ± 0.2	222 ± 2.3	10885 ± 113	8.5 ± 0.4	208 ± 1.2	10218 ± 57
9	8.5 ± 0.5	220 ± 4.4	10787 ± 216	8.6 ± 0.2	207 ± 0.6	10158 ± 30
10	10.5 ± 0.1	215 ± 7.5	10542 ± 368	10.8 ± 0.8	208 ± 0.5	10191 ± 26
11	5.7 ± 0.1	183 ± 1.3	8973 ±64	5.7 ± 0.1	205 ± 1.2	10075 ± 57
12c**	10.1 ± 0.1	155 ± 5.7	7600 ± 279	10.1 ± 0.1	144 ± 4	7058 ± 192
*****Mean ± SD**	**9.5 ± 0.2**	**213 ± 22**	**10689 ± 634**	**9.5 ± 0.4**	**203 ± 2.2**	**9961 ± 106**

** Peak force estimated using F=ma*
*** c-Control Material-PlastaZote LD-60*
**** Mean values shown only include samples 1 to 11 (control removed from calculations)*

Some samples showed signs of degradation/plastic (permanent) deformation on the first impact. The deformation occurred at the area of contact between the anvil and the impactor (Figure 3.6)

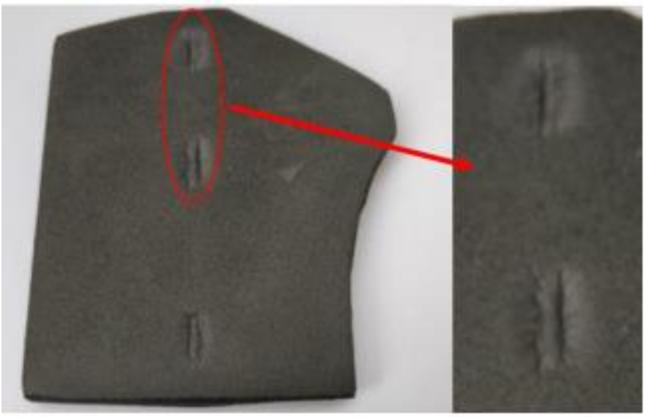


Figure 3.6 - Manufacturer sample with areas of plastic deformation highlighted in red.

3.3.3 Discussion of the Replication of Test Procedures

After a critical review of Regulation 12 and completion of the impact testing presented in sections 3.3.1 and 3.3.2 many potential concerns regarding Regulation 12 arise. A summary of the key issues are outlined below:

Impact test setup does not represent a rugby impact

- The mean peak force of the impacts (10432 N) is far greater than the impact force range of a rugby tackle (2000 – 6000 N) found by Seminati et al. [62].
- Both the striker and shoulder surrogate are unrepresentative of the body part they are embodying, mainly due to their rigid nature. This can cause differences in:
    - The magnitude and rate of deformation of the body segment (i.e. shoulder/ thigh).
    - The magnitude of stress and strain in the body segment.
    - The extent of damage that may be caused to the body segment.
    - The proportion of strain energy absorbed by the shoulder padding and body segment.
    - The interaction of the shoulder padding with the body segment and subsequent distribution of pressure and magnitudes of stress and strain experienced by the shoulder padding.

Impact test protocols make repeatability of testing problematic

- Plastic deformation of the sample was sometimes seen, the material would therefore not recover and the paddings force attenuative properties altered.
- During the 50°C test the padding decreased in temperature from impact 1 to 3. Mean sample temperature decreased from 45.2°C (impact 1) to 32.1°C (impact 2) to 28.5°C (impact 3). There is no requirement in Regulation 12 to re-heat the material after each impact and therefore a temperature decrease occurs during testing.

Performance requirements

- The minimum peak acceleration (150g) value has no research to back it up.
- Not having an upper limit can allow a manufacturer to apply minimal padding to a jersey with low/ no force attenuation properties. Branding the product 'padded clothing' could therefore be misleading to the consumer.
- Testing both sides of the padding will also give insight into whether padding protects the wearer or increases the risk of injury to the opponent.

Density measurement

- Density is the only physical property controlled by the regulation and the padding can be hard to measure. Technologies may be developed that can cause harm to players because of an increased stiffness but also pass density regulations. Using a parameter like stiffness could help to restrict materials that could cause harm to a player.

3.4 Formulation of Rugby Impact Model

To establish the specific requirements for improved testing of rugby shoulder padding using a shoulder surrogate, a shoulder impact model specific to rugby must be formulated. This model must be comprehensive and clear to consider all factors that may influence both a shoulder impact and injury. Recent research by Payne [168] which involves the fabrication of a thigh surrogate developed a deterministic contextual sequential (DCS) model what can be applied to a sporting injury situation. The rugby impact model presented uses this as a guideline and applies it to a shoulder impact in Rugby Union.

3.4.1 Generalised Model

The DCS model developed by Payne [168] uses a core framework that can be applied to many sporting impact situations. It highlights how certain elements which relate to the context, physiology, and mechanics of an impact event can be linked, this is presented in a simple flow chart. Figure 3.7 illustrates this core framework.

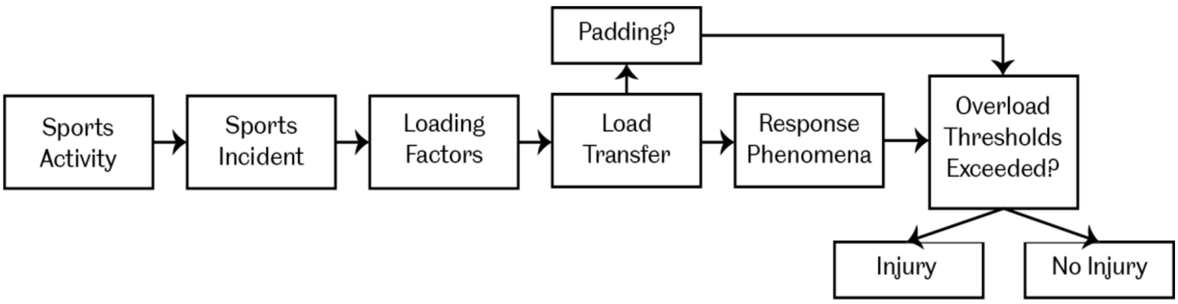


Figure 3.7 – DCS core framework (Adapted from Payne [168]).

Each element of the above flow chart is described below; this shows how each element links to another:

1. **Sports Activity** – The sport being performed when the impact occurs, relating to the environment, regulations of the sports, and style of play.
2. **Sports Incident** – The parameters of the surrogate and the striking object before contact, this relates to the objects involved, constraints, and locations.

3. **Loading Factors** – The kinetics, kinematics, and parameters of the impacted body and striker at point of impact.
4. **Load Transfer** – The conditions experienced by the impacted body during this Impact, relating to its size and material properties.
5. **Padding** – The padding or PPE worn by the surrogate or athlete.
6. **Response Phenomena** – The loading response presented by the surrogate at impact.
7. **Overload Thresholds Exceeded** – Does the stress and strain experienced by the impacted body exceed its tolerance.
8. **Injury** – If tolerances are exceeded an impact will occur.

3.4.2 Applied Model

In rugby, many impacts to the shoulder are caused when a tackler attempts to tackle a ball carrier; A similar movement is seen when a player joins a ruck or a maul. These impacts can cause various injuries to the shoulder if its injury threshold is exceeded. Some assumptions must be made due to the nature of Rugby Union. The in-game actions described above are player-on-player contacts, however, the exact area where contact is made by the shoulder varies. Correct tackle technique will however see the shoulder make contact with the thigh of the ball carrier, this is, therefore, the most common and should be used in a shoulder impact framework.

Table 3.2 - Shoulder impact of tackler in rugby.

	Rugby shoulder impact – tackler
Sports Activity	1.1. Contact sport, with on average 457 impacts per match [177]. 1.2. Player exposure: up to 80 minutes a game, normal training week for in a professional environment is 1 game and 4 training sessions a week. 1.3. Single sex sport played by males and females separately. 1.4. Team sport played by 15 players (a team) or 7 dependent on format. 1.5. Considered an aggressive contact sport. 1.6. Tackler may use a range of styles to make tackle. 1.7. Normal season runs from September to May meaning environment can change drastically.
Sports Incident	2.1. Human to human impact. Shoulder of tackler is the striker, constrained by mass and player contact with the floor. 2.2. Target is tackled player, legally must be below shoulders, usually on thigh of the tackled player.

Loading Factors	3.1. Tackler shoulder velocity averages 5.6 m/s and ball carrier 4.8 m/s.[56]. 3.2. Average mass of professional rugby player is 110.6 kg (forwards), 91.4 (backs) [178]. 3.3. Striker geometry is shoulder shape with target geometry being variable. 3.4. Striker and target mass made up of human soft tissue.
Load Transfer	4.1. Segment mass of both shoulder and target 4.2. Segment area of both shoulder and target 4.3. Material properties of human tissues. 4.4. Muscle relaxed or unrelaxed
Padding	5.1. Shoulder padding worn or not worn on shoulder of tackler. 5.2. Padding up to 12mm, density up to 60 kg/m^3
Response Phenomena	6.1. Impact duration. 6.2. Displacement of human tissue. 6.3. Peak acceleration of striking shoulder and target. 6.4. Impact force seen at the shoulder.
Overload Thresholds Exceeded	7.1. Compressive stress on skin, muscle, bone of the shoulder. 7.2. Shear stress on skin, adipose and muscle tissue. 7.3 Age, Level of fitness and repeated injuries can alter the threshold.
Injury	8.1 Lacerations 8.2 Abrasions 8.3 Contusions 8.4 Shoulder joint injuries (strained ligament, dislocations) 8.5 Fractured bone i.e., Clavicle

The applied model in Table 3.2 can be used to guide laboratory reconstructions, computer models, and applied gameplay scenarios of shoulder impacts in rugby. These can then be measured, controlled, and altered to achieve a scenario more representative of a real game environment and therefore improve testing and performance assessment of shoulder padding. What can be seen from Table 3.2 is that there are some aspects relating to the loading factors and load transfer of a rugby shoulder impact that are not easily quantifiable, possibly because of how variable these can be.

3.5 Perceptions and Attitudes Towards Shoulder Padding and Shoulder Injury in Rugby Union

3.5.1 Overview

This section has been adapted from a published submission to the Journal of Science in Sports and Exercise.

Hughes, A., Carré, M., & Driscoll, H. (2022). Perceptions and attitudes towards shoulder padding and shoulder injury in rugby union. Journal of Science in Sport and Exercise, 4(1), 66-73.

The roles of the other authors for this paper in relation to the project are as follows: M. Carré (supervisor: academic) H. Driscoll (supervisor, academic). The Manuscript was written by A. Hughes, all authors commented on the manuscript. 100% of the research and 100% of the writing was done by A. Hughes.

ABSTRACT

Purpose: To develop an understanding of the role of shoulder padding in rugby by investigating player perceptions and attitudes towards shoulder padding and extending research into shoulder injuries in rugby. **Methods:** An online survey was distributed to past and current rugby players over 13 years old in 2018. Questions related to the participants' demographic, attitudes to shoulder padding and shoulder injury history. **Results:** 616 rugby players responded to the survey. 66.1% of respondents had worn shoulder padding at some point. The age group 24-29 (ΔR^2 = 0.03, B = -0.53, p = 0.015) had an inverse association with padding effectiveness while playing experience groups 1-2 years (ΔR^2 = 0.03 B = 0.8, p = 0.032), 3-5 years (ΔR^2 = 0.03, B = 0.70, p = 0.002) and 6-9 years (ΔR^2 = 0.03, B = 0.41, p = 0) had a positive association. 37.1% of respondents considered shoulder padding to be effective at preventing Cuts and Abrasions with 21.9% finding it very effective. 50.3% considered it to be effective or very effective (9.7%) at preventing Contusion. 45.5% wore padding for injury prevention, while 19.2% wore padding to protect from reoccurring injury. Sprain/ ligament damage (57.5%) and bruising (55.5%) were the most commonly reported injuries.
Conclusions: The primary reason for wearing shoulder padding was as a means of injury prevention. Research should focus on quantifying the injury preventive capabilities of shoulder padding. Bruising, Cuts and Abrasion injuries to the shoulder are prevalent presenting new findings that these injuries are underreported.

3.5.2 Aims and Objectives

The study seeks to develop an understanding of player perceptions and attitudes towards shoulder padding as well as extend previous research regarding shoulder injury in rugby. First, the study aims to develop detailed knowledge of players' attitudes and perceptions of shoulder padding through a mixed-methods design, while examining how different subgroups may differ in their perceptions and attitudes. Secondly, the study aims to examine shoulder injury epidemiology of rugby players, including any effects of players' attitudes and perceptions of shoulder padding.

3.5.3 Methods

Survey Development

After institutional ethical approval an online survey was developed. During the preparation of this study, 25 rugby players contributed to the development of the survey through commenting on an initial set of pilot questions. After evaluation of this pilot via interview with pilot testers a final questionnaire was presented as an online survey using Google Forms.

Section 1 of the survey collected demographic and playing information. Section 2 then collected participants' attitudes and perceptions to shoulder padding, these were based on previous research relating to headgear [179], and included questions regarding shoulder padding usage, reasons for wearing and not wearing shoulder padding using open ended text box style questions, as well as participants' perceptions of how effective shoulder padding is with regards to injury prevention both generally using a 5-point Likert scale (1='not at all', 5='a great deal') and specifically to certain injuries using a different 5-point Likert scale (1='very ineffective', 5='very effective'). Injuries were grouped by type based on previous rugby based consensus statements Fuller [22] and are as follows (Cuts and Abrasions, Bruising, Sprain/ Ligament damage, Nerve injury, Dislocations, Bone injury), examples of each were given on the survey. Section 3 then collected information regarding the participants' shoulder injury history to date so that shoulder pad usage and attitudes could be linked with shoulder injury experience as well as add to epidemiological data. Participants were asked to recall their career injury history and categorise them into the previously mentioned categories, no other injury history information was taken due to the possibility for systematic error. The questionnaire included both closed and open questions. This mixed methods design allowed for descriptive and interpretive information to be obtained.

Survey Deployment

Rugby players aged 13+ of any gender and skill level were targeted during the deployment of the questionnaire, parental consent (under 18s) was taken. The questionnaire was distributed to respondents between May and July 2018. The questionnaire was publicised through various social media platforms including directly through World Rugby's twitter handle. Various English rugby clubs were also approached, and the survey link was sent to its members. The country in which the respondents resided was not controlled and was only available in English.

Data Analysis

Quantitative data was inputted into SPSS (version 25) and descriptive statistics were produced in order to examine demographics, shoulder pad usage, and shoulder injury history. Any incomplete data was disregarded. After parametric checks, ordinal regression analysis was performed to identify significant predictors for two dependent variables, (the perceived effectiveness of padding and specific injury history i.e. dislocations, bruising). Open ended survey responses (reasons for wearing and not wearing shoulder padding) were examined using a thematic approach, as used by Braun and Clark [180]. Eight higher order themes were identified for the open ended questions using an inductive approach. Raw data was coded into groups by the principal researcher, this process was then discussed with the research team and a consensus made to ensure trustworthiness of the data. Descriptive statistics for these themes were then produced in order to examine the responses.

3.5.4 Results

Basic Characteristics

At total of 616 responses were collected from the survey, giving a wide demographic of rugby players (Table 3.3).

Table 3.3 - Demographic information of players surveyed.

Characteristic		**Responses (number, (%))**
Sex	Male	574 (93.2)
	Female	40 (6.5)
	Prefer not to say	2 (0.3)
Age	13-17	33 (5.4)
	18-23	217 (35.2)
	24-29	146 (23.7)
	30-35	82 (13.3)
	36+	138 (22.4)
	Under a year	10 (1.6)

Playing Experience	1-2 years	25 (4.1)
	3-5 years	87 (14.1)
	6-9 years	104 (16.9)
	10+ years	390 (63.3)
Highest Playing Level	School	10 (1.6)
	Junior Club	28 (4.5)
	Junior County	10 (1.6)
	Academy	18 (2.9)
	University	112 (18.2)
	Senior Social	115 (18.7)
	Senior Amateur	255 (41.4)
	Semi-Professional	57 (9.3)
	Professional	10 (1.6)
Playing Position	Front Row Forwards	182 (29.5)
	Back Five Forwards	223 (36.2)
	Backs	211 (34.3)

Shoulder pad use

66.1% (n=407) of players had worn shoulder padding at some point. 9.9% (n=61) always wore shoulder padding, 17.7% (n=109) only wore shoulder padding during matches, 13.1% (n=81) wore shoulder padding, but only because of an injury and 25.3% (n=156) wore shoulder padding regularly in the past but at present did not. 61% (n=111) of front row forwards, 61% (n=136) of back five forwards and 74% (n=129) of backs had worn shoulder padding at some point.

Attitudes towards effectiveness of shoulder padding

The median perception of the effectiveness (Likert scale 1-5) of padding was 2 (Inter Quartile Range (IQR) = 2-3). When player's behaviours were factored in, the results were, those that always wore shoulder padding (Median = 3, IQR = 3), only wore shoulder padding in matches (Median =3, IQR = 2-3), wore shoulder padding, but only because of an injury (Median = 2, IQR = 2), wore shoulder padding regularly in the past but at present did not (Median = 3, IQR = 2-3), and had never worn shoulder padding (Median = 3, IQR = 2-3). Based on the regression model, those that always wore shoulder padding (ΔR^2 = 0.19, B = 2.25, SE = 0.28, CI 1.70 – 2.80, p = 0), only wore shoulder padding in matches (ΔR^2 = 0.19, B = 1.81, SE = 0.23, CI 1.36 – 2.23, p = 0), and those that wore shoulder padding, but only because of an injury (ΔR^2 = 0.19, B = 0.59, SE = 0.24, CI 0.12 – 1.06, p = 0.014) had a positive association with perceived effectiveness. Gender (p=0.245), playing position (p=0.109) and playing level (p=0.540) had no significant association with the perceived effectiveness of padding. However, when taking into account age, the group 24-29 had an inverse association with shoulder padding effectiveness (ΔR^2 = 0.03, B = -0.53, SE = 0.22, CI -0.95 – -0.10, p = 0.015). Playing experience groups 1-2 years (ΔR^2 = 0.03 B = 0.8, SE = 0.04, CI 0.07 – 1.53, p = 0.032), 3-5 years (ΔR^2 = 0.03, B = 0.70, SE = 0.22, CI 0.26 – 1.13, p = 0.002) and 6-9 years (ΔR^2 = 0.03, B = 0.41, SE = 0.2, CI 0.02 – 0.80, p = 0) had a positive association with padding effectiveness.

Respondents considered shoulder padding to be either effective (37.1%) or very effective (21.9%) at preventing Cuts and Abrasions. 50.3% considered it to be effective and 9.7% very effective at preventing bruising. 17.4% of respondents considered it either effective or very effective at preventing sprain/ ligament damage, as well as 10.6% for dislocation and 21.5% for bone injury (Figure 3.7). Based on the regression model, whether a player had received a specific injury had no association with their perceived effectiveness of shoulder padding except for a bone injury. A positive association was found between perceived effectiveness of shoulder padding preventing bone injury (ΔR^2 = 0.01, B = 0.50, SE = 0.21, CI 0.095 – -0.9, p = 0.016) and whether a player had received a bone injury as a result of playing rugby. This indicating that players who had received a bone injury thought shoulder padding was more effective at preventing this injury than player who had not received a bone injury.

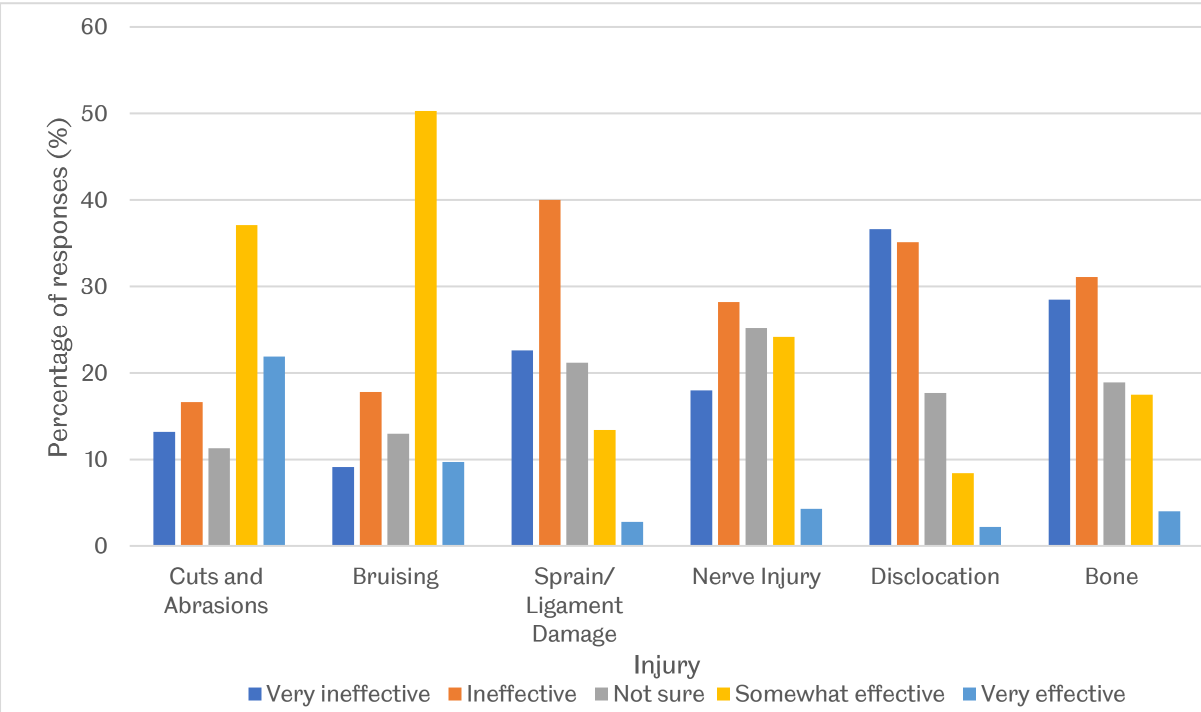


Figure 3.7 - The perceived effectiveness of padding for specific injuries.

Attitudes of players who wear shoulder padding

Eight themes were identified when considering players who had worn shoulder padding at some point (Table 3.4). Of these players, 62.6% of responses indicated wearing shoulder padding as a form of protection or injury prevention with 19.2% of these being to protect from a reoccurring injury. 15.8% of responses implied rugby players wore shoulder padding to feel more confident, mainly in the tackle situation. 9.3% of responses indicated wearing shoulder padding for comfort in impacts rather than as a form of protection.

Table 3.4- Reason themes for wearing shoulder padding (listed from most to least common).

Higher order themes (n=386)	Example Responses
Injury Prevention and Padding (43.5%)	Protection. Protect from minor shoulder injury. Degree of protection offered to shoulder and collar bone in contact. Protect against soft tissue injury.
Protection from reoccurring injury (19.2%)	To protect my shoulder whilst it wasn't 100%. Returning from an injured shoulder. To reduce impact on shoulders following an injury. Damaged my ACjoint and padding it was the only way I could tackle with the least amount of discomfort.
Confidence (15.8%)	When I first played contact rugby, it gave me greater confidence when making a tackle. Confidence in the tackle area. Purely confidence. I don't believe it helps, other than my mind. Feel more secure. It makes me feel more confident about making tackles in matches.
Comfort in impacts (9.3%)	Just gives a little bit of extra comfort in the pack for tacking and scrums. Less sore shoulders after scrum. Gives me more comfort when making tackles on oppositions bony parts.
Recommendation from coaches, friends or parents (7.3%)	When I was younger I wore it for shoulder protection mainly on the insistence of my Mum. Was recommended by the coach. It was popular to wear them.
Habit (1.8%)	It feels part of my gear, same as gumshield, shorts etc. Was given to me for free, got used to wearing it and then didn't like the feel of playing without it.
To change own physical appearance (1.6%)	Being smaller than everyone else. Due to my size frame shoulder pads helped make me feel bigger, it had a bit of placebo effect.
To try it out (1.6%)	No specific reason, a friend gave it to me and I decided to try it out.

Attitudes of players who do not wear shoulder padding

Eight themes were identified when considering players who did not wear or chosen to stop wearing shoulder padding (Table 3.5). 38.6% of responses indicated wearing shoulder padding was not needed, with 21.3% of responses indicating shoulder padding was uncomfortable. 16.8% of responses indicated rugby players did not feel padding had added protective benefits

Table 3.5 - Reasons for not wearing shoulder padding.

Higher order themes (n=352)	Example Responses
They are not required (38.6%)	I stopped wearing it as I didn't need them to absorb impacts anymore. Just never bothered with it. I don't see the need for shoulder padding, I've never hurt my shoulders before. Injury healed so no longer required shoulder pad protection.
Discomfort (21.3%)	I stopped as it was uncomfortable and I tended to overheat. Can get too hot wearing them and sometimes uncomfortable. I get too hot wearing them otherwise I would probably wear them all the time. I feel claustrophobic in them at times and get too hot.
Do not offer protection (16.8%)	I am unaware of the difference it could make to my safety or skills. Didn't seem to help with anything as so thin. No added benefits to protection.
Restricts movement (6.3%)	It adds bulk, makes it harder to manoeuvre. Movement limiting. My movement felt restricted with the pads, and I wanted full movement to avoid injury.
Cost and Availability (6.3%)	It seems unnecessary and is an expense I can't really afford. Too costly to replace.
Impacts the game negatively (4%)	I enjoy the hard-hitting nature of the game which I feel would lack with pads. Not wearing shoulder padding encourages a correct technique in tackle/contact situations and observation of the laws of the game. Wearing padding too easily encourages reckless and undisciplined hits from bad angles with greater force. Enjoying the tackle more without them.
Stigma (3.7%)	Not the manly thing to do. It's for girls. There is a perception of people who wear padding being 'soft'.
False sense of security (3.1%)	It gives a false sense of security, if you're going to break your bones, you're going to break your bones. Disagree with it. I believe it gave a false belief to those who did.

Shoulder injury data

72.8% (n=447) of players reported a shoulder related rugby injury. Of those that reported having a shoulder related injury, 35.8% (n=160) reported experiencing a Cut or Abrasion injury, 55.5% (n=248) a bruising injury, 57.5% (n=257) a sprain/ ligament related injury, 33.1% (n=148) a nerve related injury, 18.1% (n=81) a dislocation and 20.0% (n=89) a bone related injury. Using the regression model, players use of padding could be used to predict what specific injuries they had sustained, the category 'I've worn shoulder padding, but only because of an injury' was discounted. Players that always wore shoulder padding (ΔR^2 = 0.033, B = 0.63, SE = 0.30, CI 0.05 – 1.21, p = 0.034), only wore shoulder padding in matches (ΔR^2 = 0.033, B = 0.48, SE = 0.25, CI 0 – 0.96, p = 0.049), and have worn padding regularly in the past but at present do not (ΔR^2 = 0.033, B = 0.46, SE = 0.22, CI 0.03 – 0.89, p = 0.038), had a positive association with having sustained a bruising injury. Players that only wore shoulder padding in matches (ΔR^2 = 0.04, B = 0.83, SE = 0.29, CI 0.27 – 1.39, p = 0.003), and have worn padding regularly in the past but at present do not (ΔR^2 = 0.04, B = 0.78, SE = 0.26, CI 0.26 – 1.29, p = 0.003), had a positive association with having sustained a sprain/ligament injury. Players that only wore shoulder padding in matches (ΔR^2 = 0.07, B = 0.99, SE = 0.34, CI 0.32 – 1.67, p = 0.004), had a positive association with bone injury. Figure 3.8 displays specific shoulder injury history as a function of shoulder padding usage. Backs sustained less shoulder injuries (66%), when compared to front row forwards (79%) and back five forwards (74%). 89% of the front row that always wore padding had sustained an injury compared with the 66% that had never worn padding. However, 50% of the backs that always wore shoulder padding had sustained a shoulder injury, this was the same for the backs that never wore padding (50%).

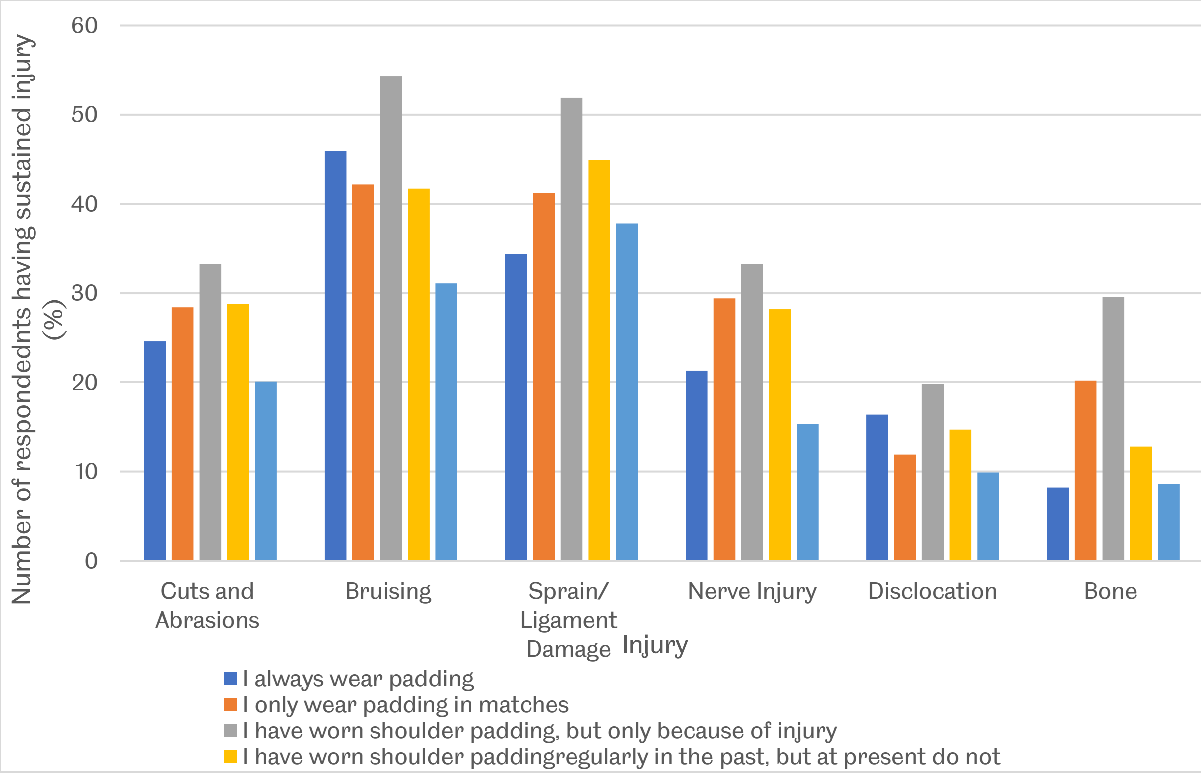


Figure 3.8 - Specific shoulder injury history as a function of shoulder pad usage.

3.4.5 Discussion

Shoulder Padding

The regression model showed increased perceived effectiveness of padding with increased use. Both players who always wore padding, only wore padding in matches and those that wore shoulder padding, but only because of an injury had a positive association with the perceived effectiveness of shoulder padding. When exploring this further, both the variables age and playing experience influenced perceived effectiveness of shoulder padding. The age group 24-29 had an inverse association with perceived effectiveness and playing experience groups 1-2 years, 3-5 years and 6-9 years had a positive association. It would be very exploratory to state a reason for this, however it is suggested male rugby players are at their peak muscle mass in the 24-29 age group, therefore may feel they do not need shoulder padding as a result [181]. When taking demographic information into account, no other group had a significant positive or negative association with perceived shoulder padding effectiveness. Whilst there seems to be a good awareness into the limitations shoulder padding has at preventing injury, further education should be directed to all playing groups in order to reinforce player knowledge.

59% of respondents considered shoulder padding to be either effective or very effective at preventing Cuts and Abrasions and 60% of respondents considering shoulder padding to be either effective or very effective at preventing bruising injury, complimenting previous research into padded headgear, finding 55% of respondents to consider headgear to be effective at preventing minor injuries [179]. Shoulder padding's ability to reduce the risk of superficial injuries like Cuts and bruising must be measured in order to justify rugby players' perceptions of padding. 10.6% considered shoulder padding to be either effective or very effective at preventing dislocations, as well as 21.5% considering shoulder padding to be either effective or very effective at preventing bone injury. However, this is yet to be proven or quantified, this also does not align with World Rugby's™ views. Further education as well as responsible marketing from manufacturers and governing bodies should be considered to ensure fewer rugby players view shoulder padding as an effective tool at preventing severe injuries.

The primary reasons for wearing shoulder padding were as a means of injury prevention (43.5%) or to protect from reoccurring injury (19.2%). This was expected due to how shoulder padding is commercially branded and its proven impact force attenuating abilities [182]. 15.8% of players wore shoulder padding to increase confidence, mainly in the tackle. The outweighing association that players use shoulder padding as a means of injury prevention suggests this increased confidence stems from a decreased worry about getting injured. This result is similar to a study by Barnes et al., [179] on protective rugby headgear where 13% of responses related to increased confidence as a motivation for its use. It is however, important to note World Rugby™ does not view shoulder padding as a form of protective equipment and has set impact attenuating abilities to a maximum limit, with the view of not over protecting players and changing their on pitch behaviours [1]. It has however been suggested that some players can become overly reckless when wearing protective equipment [183], further backed up by 3.1% of reasons for not wearing shoulder padding being related to the feeling of a false sense of security.

The primary reason for not wearing padding was that shoulder pads were not needed in rugby (38.6%). Previous research suggests the physical nature of the game leads to players adopting a mind-set where extra padding is not needed [184]. Discomfort (21.3%) and the feeling of restricted movement (6.3%) were also key reasons for not wearing padding. Similar to research into padded headgear in rugby, which also found discomfort and heat regulation issues to be primary reasons for not wearing padded headgear [185]. 16.8% of respondents felt shoulder padding offered no extra protection. Further research into what injuries shoulder padding may reduce the risk of is needed followed by education of these findings to rugby players. Manufacturers should consider the factors of discomfort and restricted movement while also acknowledging World Rugby™ regulations when designing future products.

Shoulder Injury

Sprain/ ligament damage (57.5%) and bruising injuries (55.5%) were the most prevalent. Previous research reports a lower frequency of bruising injuries (12 – 17% [63, 64]). Possibly due to the injury definition used in both studies (24 + hours' time loss) which would lead to the underreporting of a bruise that may not be of the severity to cause time loss or require medical attention. As well as this, it is possible players in the current study were more likely to respond to the survey if they had had a shoulder injury. Comparing this data to the data mentioned must be done with caution due to the significant differences in approaches taken. The large prevalence of reported bruising injuries to the shoulder does suggest shoulder padding's ability to decrease the risk of a bruise should be explored. This also the case with Cuts and Abrasion injuries, 35.8% of respondents had sustained a Cut or Abrasion as a result of playing rugby. No published research reports Cuts, Lacerations, or Abrasions specifically to the shoulder region. With regards to less severe injuries, players that always wore padding had sustained more Cuts and Abrasions (24.6%) and bruising injuries (45.9%) than that of players that had never worn padding (20.1%, 31.1%). Players that had never worn padding felt they did not see the need to wear it, potentially because they did not need the added protection (i.e. increased muscle mass), therefore potentially explaining the larger reporting of less severe injuries in players that always wear padding. Coupled with this, some players that had never worn padding did so out of stigma. The stigma of wearing padding may also have led to the under reporting of less severe injuries like Cuts, Abrasions and bruising.

Limitations

Limitations stem from the method of data collection, recall bias may have been an issue due to the self-reporting style of data collection, enhanced when asking participants about their non-severe injury history beyond a year [186]. There is mixed findings on the validity of self-reporting injury data in this way [187-189], future studies should use injury data reported by medical professionals. The varied demographic of respondents would have reduced selection bias, however 72.8% of respondents had had a shoulder injury, suggesting that individuals with previous shoulder injuries were more likely to respond to the study. Due to the data collection procedures of the study the severity of reported injuries was not recorded, this should be explored in the future. Future studies should explore whether shoulder pad use affects actual playing behavior as well as shoulder injury occurrence. There was limited heterogeneity in gender and playing level, with only 6.5 % of respondents being female and 10.9% semi-professional or professional, this could reduce the variability in the results. Finally, the study did not account for the nationality or region of the participants, attitudes can differ by region limiting the ability of the study to know where the data generalises to.

Conclusions

The primary reason for wearing shoulder padding was as a means of injury prevention. Research should focus on quantifying the injury preventive capabilities of shoulder padding. Bruising, Cuts and Abrasion injuries to the shoulder are prevalent presenting new findings that these injuries are underreported.

3.5.6 Summary

There is a lack of research concerning rugby player's attitudes towards padded clothing. The survey adds a substantial amount to this knowledge, to increase understanding of shoulder injuries in rugby and address RQ1 and RQ2. The following actions in turn guided the PhD project.

- The ability of shoulder padding to prevent or reduce the severity of specific blunt force trauma injuries (Contusion, Lacerations, broken bones) must be quantified to support and educate the motives behind wearing shoulder padding.
- Discomfort and restricted movement stopped players wearing padding, future materials and technologies should be explored to accommodate this market need while taking into account World Rugby's™ guidelines to limit the protection in rugby.

3.6 Targeted Research Approach

Both the rugby impact model, review of Regulation 12, and the survey informed a series of requirements for an improved Regulation 12. The key aim is to develop an impact test more representative of what happens 'on field' in a rugby shoulder impact. After this, recommendations for an improved Regulation 12 can be made. Incremental progressions from current test protocols are necessary to improve knowledge of both human impact response and shoulder padding effectiveness and therefore take steps to a more complex biofidelic shoulder surrogate. For each progression of surrogate complexity and improved test methods three key elements have been identified as a crucial part to an improved Regulation 12.:

- Simplification of Geometry and Anatomy
- Materials
- Validation Practices

3.6.1 Simplification of Geometry and Anatomy

The balance between anatomical biofidelity and repeatability, ease of fabrication, and technological boundaries must be considered. The appropriate anatomical simplifications must be considered, the assessment of human and animal tissue geometries as well as the use of FE models (PhD A – MMU) as a design tool will inform this balance. These considerations will ensure a replicable shoulder surrogate that provides a similar response to the human shoulder while also being applicable to Regulation 12 test procedures is fabricated.

3.6.2 Materials

The materials used in the shoulder surrogate are critical to its mechanical response to load. There are many tissues and structures that will affect the response of the human shoulder. The surrogate will look to mimic each structure in a multi-layer approach. The surrogate must have an appropriate level of complexity and a pragmatic balance between biofidelity and practically needs to be considered. The assessment of the mechanical properties of each tissue must be considered, these material properties can be identified both through the use of previous research as well as the measurement of organic tissue *in vivo* and *ex vivo.* How these mechanical properties can be affected must also be considered.

3.6.3 Validation Practices

To develop a biofidelic shoulder impact surrogate that can be used to assess shoulder padding and improve Regulation 12 test protocols, a consistent set of validation procedures must be developed. Mechanical properties, as well as the mechanical response to impact of a shoulder surrogate, can be measured and compared to organic tissues. These stresses can be measured and shoulder paddings performance assessed.

An ideal validation procedure would be to use a living human however ethical restrictions prevent this unless performed at un-harmful loads. The next best alternatives must therefore be applied. This could be through the use of animal tissues (i.e. Porcine) with similar properties to humans.

3.6.4 Improved Regulation 12 Plan

Given each section outlined in chapter 3, an approach to the improved Regulation 12 test procedures has been formulated. The work plan and its interaction with each element are seen in Figure 3.9. Having been established, this work plan sets out the activities which will be discussed in the remainder of this thesis.

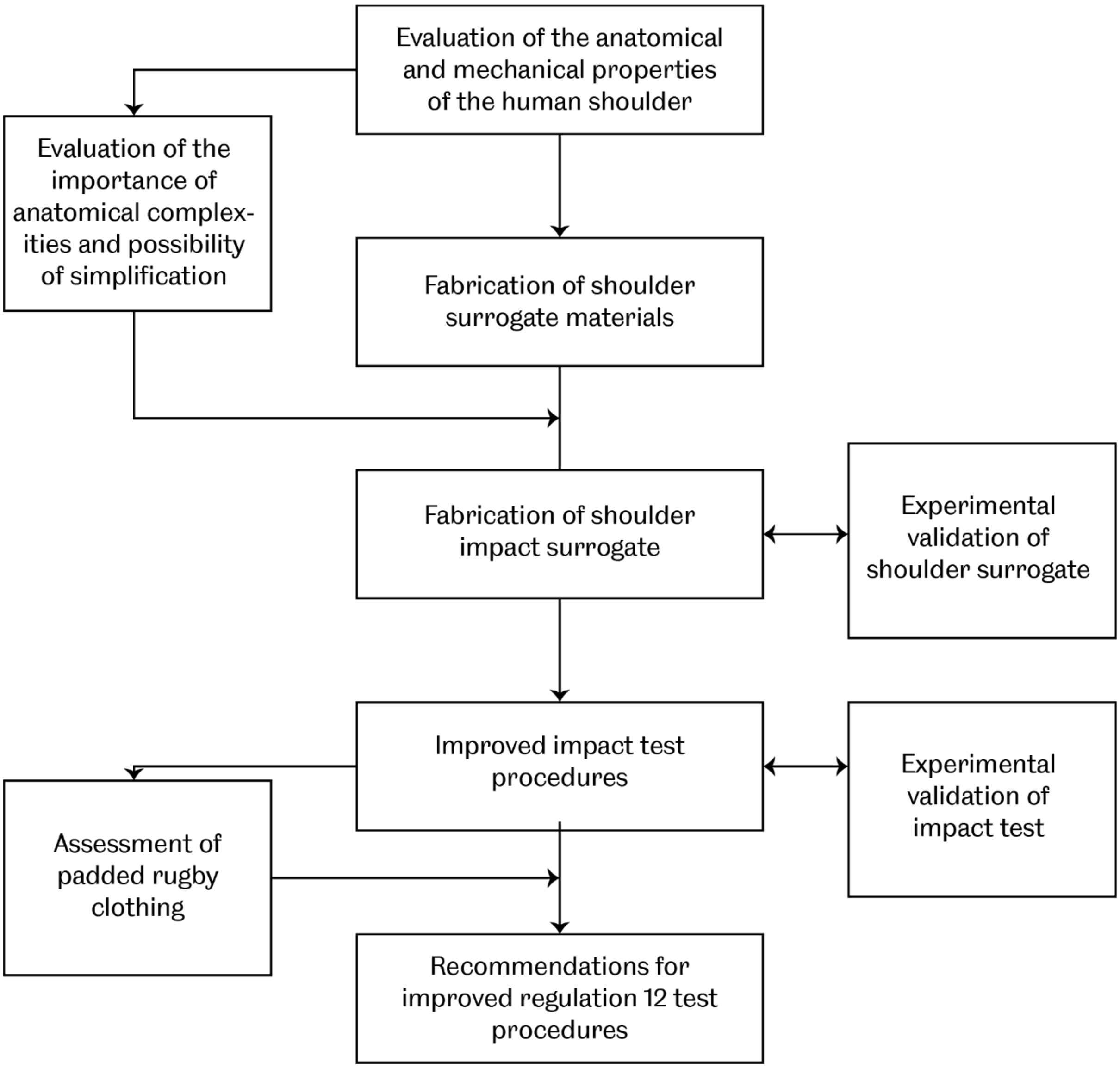


Figure 3.9 - Project workflow.

CHAPTER 4 - ANATOMICAL AND MECHANICAL ASSESSMENTS FOR THE DEVELOPMENT OF A HUMAN SHOULDER SURROGATE

4.1 Chapter Overview

This chapter details an examination of the human shoulder complex, adding to that detailed in chapter 2, including all the data collection necessary to design a human shoulder surrogate. External and internal anatomical measurements have been assessed through experimental testing to complement published datasets. The mechanical properties of organic tissues have been assessed experimentally and compared to past literature. The feasibility of simplifying these geometries for ease of fabrication and repeatability of testing for the development of a human shoulder surrogate is also discussed.

4.2 The Shoulder Complex

The shoulder makes up a large part of the upper limb attaching this body part to the torso. The shoulder joint is structurally and functionally complex as it is one of the most freely moveable areas in the human body. When considering its external anatomies, the shoulder runs at an angle (underlying is the trapezius muscle) from the neck to a flat region (underlying is the AC joint). This then drops down (underlying is the deltoid muscle) into the arm (Figure 4.1a). At the posterior of the shoulder, the external anatomies remain unchanged, with the scapula (bone) running down the back, underneath the skin (Figure 4.1b).

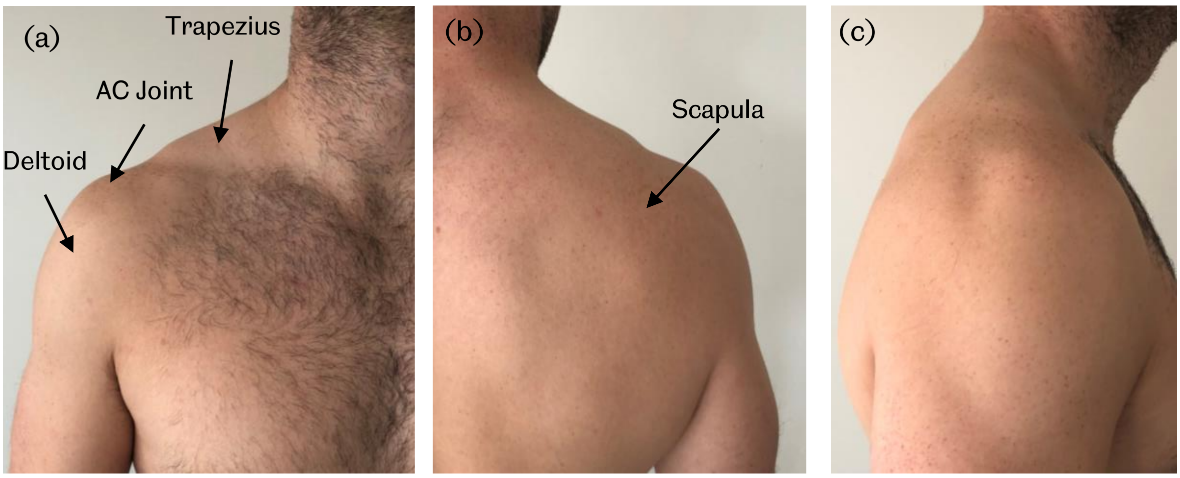


Figure 4.1 (a) – The shoulder (anterior view), (b) – posterior view, (c) – Lateral view.

4.2.1 Bones

Four main bony structures make up the shoulder and give it its rigidity (Figure 4.2):

- The Humerus attaches the arm to the shoulder.
- The Clavicle runs from the joint to the Sternum. A fractured clavicle is a common rugby injury.
- The Scapula attaches the Humerus to the Clavicle and runs down the back of the rib cage.
- The Acromion which forms the summit of the shoulder, this is a bony prominence on the Scapula that connects to the Clavicle (AC joint) that can be easily located externally.

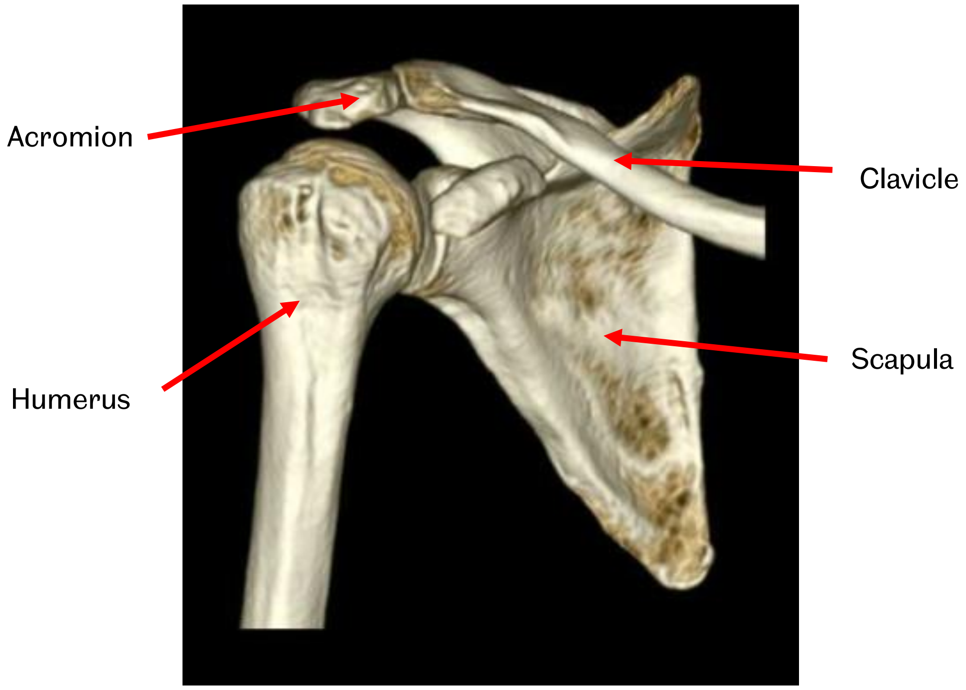


Figure 4.2 – Bones of the Shoulder Joint, anterior view (adapted from www.imaios.com).

4.2.2 Muscles, Ligaments and Tendons

A substantial amount of the soft tissue in the shoulder is muscle (Figure 4.3) The Trapezius runs from the Scapula extending longitudinaly to the spine, Making up a large proportion of the upper shoulder. The Deltoid forms around the contour of the shoulder, its origin runs from the Clavicle, Acromion and Scapula, and inserts in the Humerus. It sits laterally to the shoulder joint making up much of its mass.

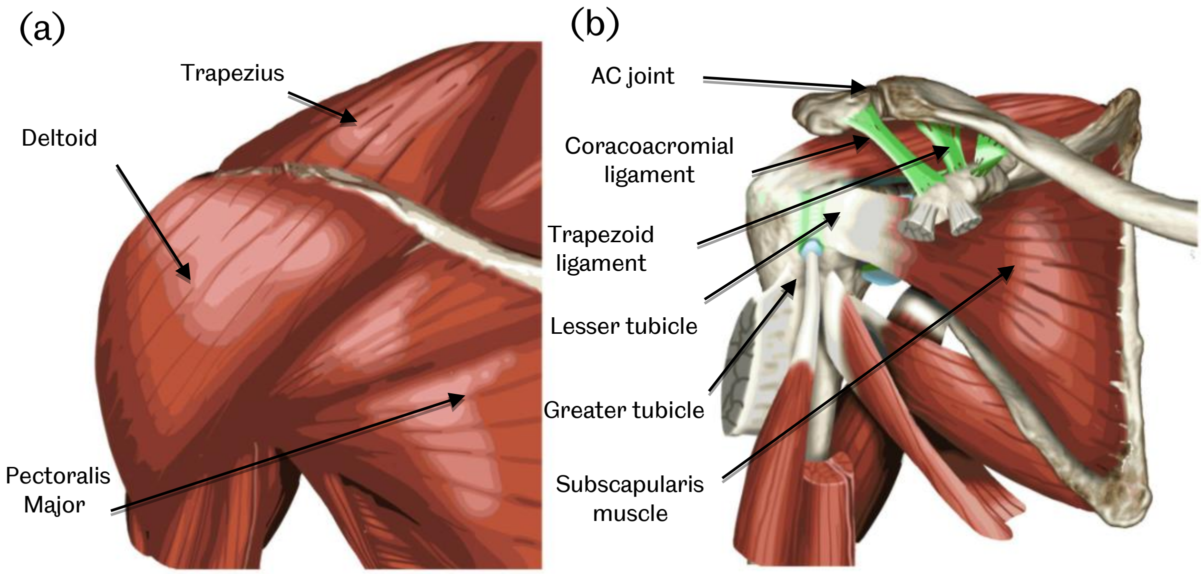


Figure 4.3 (a) – Muscles of the shoulder joint (anterior view), (b) – Ligaments and tendons of the shoulder joint (anterior view) (adapted from www.imaios.com).

Figure 4.4 shows both an illustration and an MRI scan of the shoulder cut out in a coronal view. When analysing these images it's apparent that the shoulder complex is mostly made up of bone and muscle with a small fascia layer external to the skin. When considering surrogate design, other tissues like ligaments and tendons could be removed for anatomical simplification.

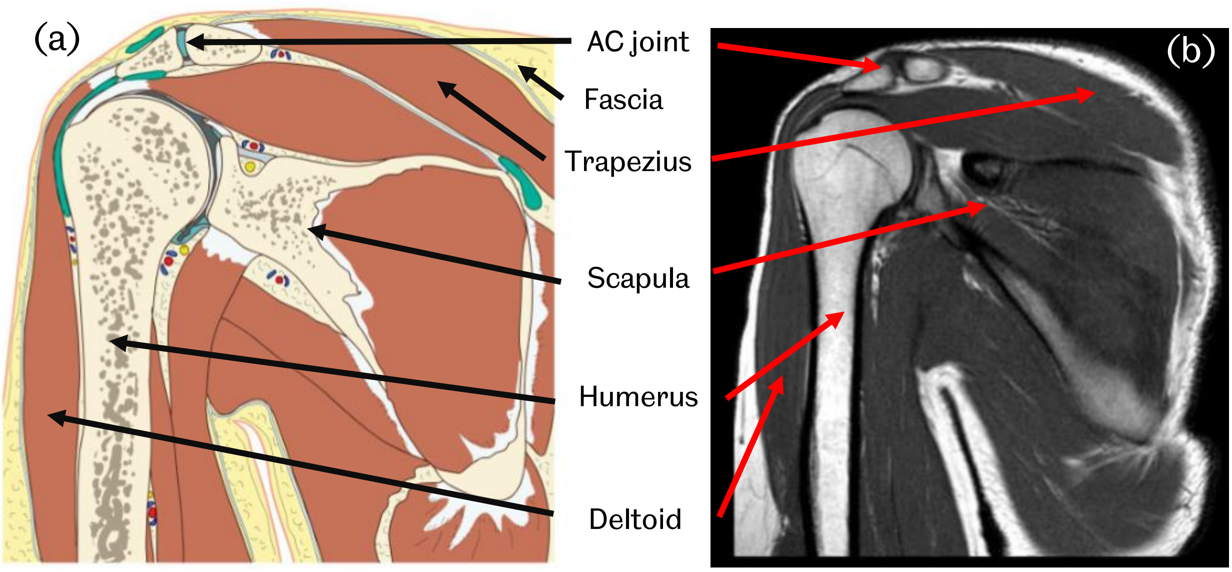


Figure 4.4 (a) - Illustration of the shoulder joint (coronal view) (b) - MRI of the shoulder (coronal view) (adapted from www.imaios.com).

4.3 External Anatomical Assessments of the Shoulder

4.3.1 3D Scanning of Rugby Players Shoulders

To establish the external shoulder geometries of rugby players' shoulders, a dataset must be evaluated. External anatomies can be evaluated according to a range of complexities. In the current study, external geometries of rugby players' shoulder segments were acquired using 3D imaging. 3D imaging systems capture detailed and accurate images of the human body, from which size and shape characteristics can be extracted. Measures obtained from 3D imaging have been used to describe, interpret, and analyse the human body in several applications, including apparel sizing, clinical evaluation, and 3D modelling of biological structures. 3D imaging technology has been used extensively within the Sports Engineering Research Group (CSER) at Sheffield Hallam University to acquire geometries of various populations; however, the collection of shoulder geometries from rugby players was novel to this project. The following explains the procedures and results of this testing.

Participants

Nine male semi-professional rugby players (National 2 or above which is tier 4 of English Rugby Union, generally this is a semi-professional level) of varying playing positions were selected for the study. Participant characteristics are shown in Table 4.1. Before testing all, participants completed an initial screening form and provided written informed consent. Players were aged 20-33 and did not have any existing shoulder injuries.

Table 4.1 – External shoulder measurement participant characteristics (n = 9).

	Mean	Standard Deviation
Age	24.3	4.27
Height (m)	1.87	0.09
Mass (kg)	102.6	9.75
BMI (kg/m^2)	29.4	2.49

Experimental Setup and Procedure

All external anatomical measurement procedures were conducted in a purpose-built human morphology lab. Participants were required to remove clothing from their upper body during measurement procedures to prevent it from masking the observed body shape. Each participant had anatomical landmark locations, which were required for 3D scan post-processing procedures, located manually through palpation and marked with a cross on the skin using a fine-tipped surgical marker (Viscot 1451). All anatomical landmarks marked during the experimental protocol are listed in Table 4.2 and shown in Figure 4.5. Calipers were used to obtain a physical measurement of the participants' half shoulder width (AC joint to a point perpendicular with the Jugular Notch) and Deltoid width. These measurements were taken with the participants in an anatomical position.

Table 4.2 – Anatomical landmarks of the shoulder region.

Anterior		Posterior	
No.	Anatomical Landmark	No.	Anatomical Landmark
1	Thyroid Prominence	6	Vertebral Prominens (C7)
2	Superior Length of The Trapezius (Front)	7	Superior Length of The Trapezius (Back)
3	Deltoid Tuberosity insertion	8	Deltoid Tuberosity insertion
4	Sternum	9	Vertebral
5	Jugular Notch	10	Acromioclavicular (AC) joint

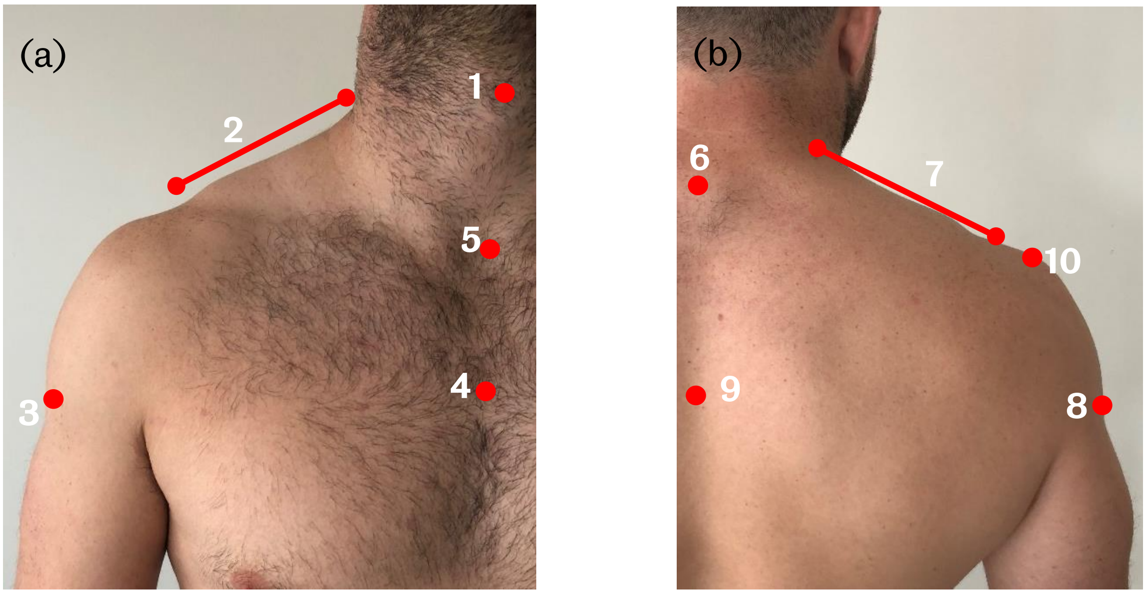


Figure 4.5 – Anatomical shoulder landmarks (a) – Anterior view of the shoulder region, (b) – Posterior view of the shoulder region.

3D imaging data of the shoulder segment was captured using an Artec Eva handheld structured light scanner (Artec, Luxembourg), with a 3D point accuracy of 0.1 mm and data acquisition speed of up to 18 million data points per second [190]. During the scans, participants were asked to bare their right shoulder and sit in an upright position with their right arm in a relaxed position and their head in a neutral position. Multiple scans were collected of each participant to ensure full coverage of the required shoulder region was captured. 3D point-cloud data collected from all scan images of each participant were initially cleaned and aligned within proprietary Artec Studio data processing software to account for subtle movements of the participant during scanning. 3D imaging data was then imported into Geomagic Studio (Geomagic Wrap 2017, Geomagic, Luxembourg) for further post-processing and modelling. Within Geomagic, each shoulder model was cleaned to fill holes and defects in the raw imaging data, and then formed into a watertight polygonal mesh and exported as a STEP file for further analysis.

Shoulder Analysis

All STEP files were imported into 3D CAD software Solidworks (Solidworks 2018, Dassault Systèmes, France) for analysis. Key geometries as shown in Figure 4.6 were then measured. These 4 measurements were: Half shoulder width, Deltoid width, AC joint to trapezium insertion, and Central Spine of the Scapula to the Central Clavicle.

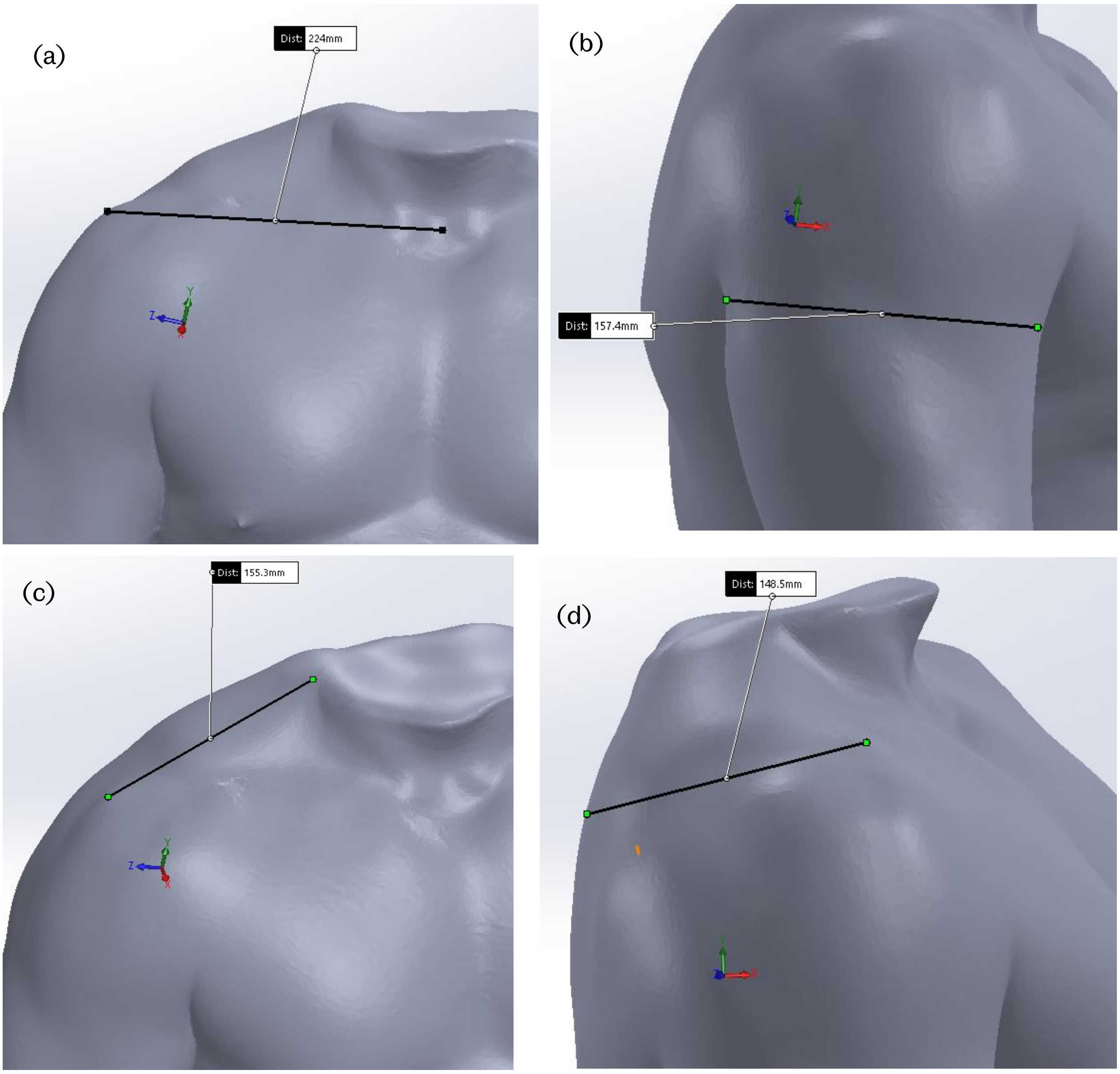


Figure 4.6 (a) – Half Shoulder Width, (b) – Deltoid Width, (c) - AC Joint to Trapezius Insertion, (d) – Central spine of Scapula to Central Clavicle.

Results

The measurements taken from the scans are in Table 4.3, when using Figure 4.6 as a guide, the shoulder approximately represents a semi-cylinder. AC joint to Trapezium insertion measurements can be used to guide the length of a shoulder surrogate, while the Central Spine of the Scapula to the Central Clavicle could guide the diameter. Full CAD images of each participant can be seen in Appendix B.

Table 4.3 – Shoulder measurements (mm), from 3D scans (n = 9).

	Mean	Standard Deviation
Deltoid Width	168	11
Half Shoulder Width	225	10
AC Joint to Trapezium Insertion	175	20
Central Spine of Scapula to Central Clavicle	155	12

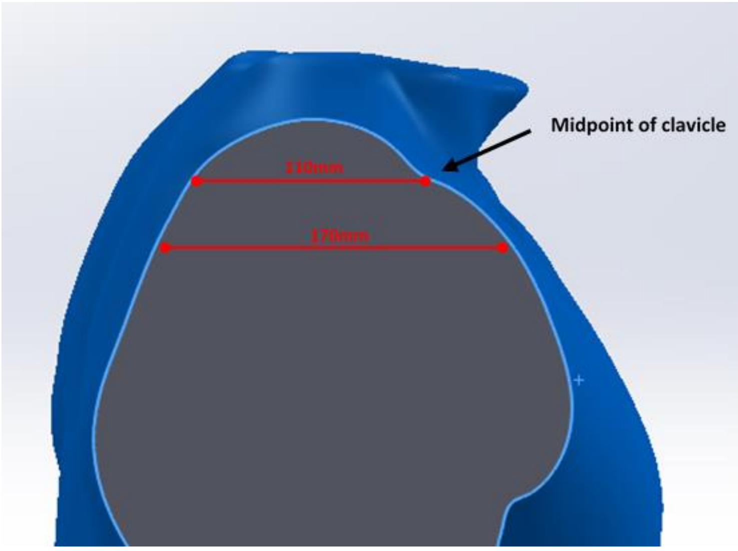


Figure 4.7 – Cross-section (from the centre of Trapezius) of shoulder scan with sagittal plane measurements.

Cross-sectional images (Figure 4.7) show the upper part (intended impact region) of the shoulder approximately represents a half-cylinder, further backing up the statement regarding Figure 4.6. In Figure 5.16, a half-cylinder geometry shoulder surrogate is overlayed on this cross-section to justify this. The dataset of rugby players' shoulders built up here can be used to guide the design of a shoulder surrogate as well as to compare with the shoulder geometry of the general population. The results from this can assist with the following considerations, as discussed further in chapter 5:

- How biofidelic should the external geometry of the shoulder surrogate be?
- How will these geometries be replicated in a surrogate mould?
- How will impact response be affected by simplification of external geometries?
- How do the results differ from other datasets?

4.4 Internal Anatomical Assessments of the Shoulder

4.4.1 Ultrasound Scanning of Rugby Players Shoulders

Section 4.2 sets out a description of the internal anatomies of the shoulder. However, to develop a shoulder surrogate these geometries must be measured using an appropriate dataset. A method to do this is ultrasound scanning, a technique that uses high-frequency sound waves whereby ultrasound images can be obtained, and tissue layers depicted. These images can then be used to measure geometries such as the thickness of layers. The following explains the procedures and results of ultrasound testing on rugby players. Like the external scanning, the right shoulder was kept in a relaxed position for this testing.

Participants

Six male semi-professional rugby players (National 2 or above) of varying playing positions were selected for the study. Players were aged 20-33, did not have any existing shoulder injuries, and were all participants in the previous 3D scanning study (§4.3.1). Unfortunately, due to participants dropping out only six were scanned. It was however important to only use participants that had participated in the previous 3D scanning study to keep a consistent dataset.

Experimental Setup and Procedure

Scanning was completed using a Telemed ultrasound system (Echo Blaster 128, Milan, Italy). Participants were sat down on a chair with their arms down by their sides in a relaxed position, and their head in a neutral position. Three landmarks on their shoulder were marked through palpitation; A midpoint between their Acromion and the seventh Cervical Vertebra on their Trapezius, AC joint, and the outermost part of the deltoid as seen in Figure 4.8.

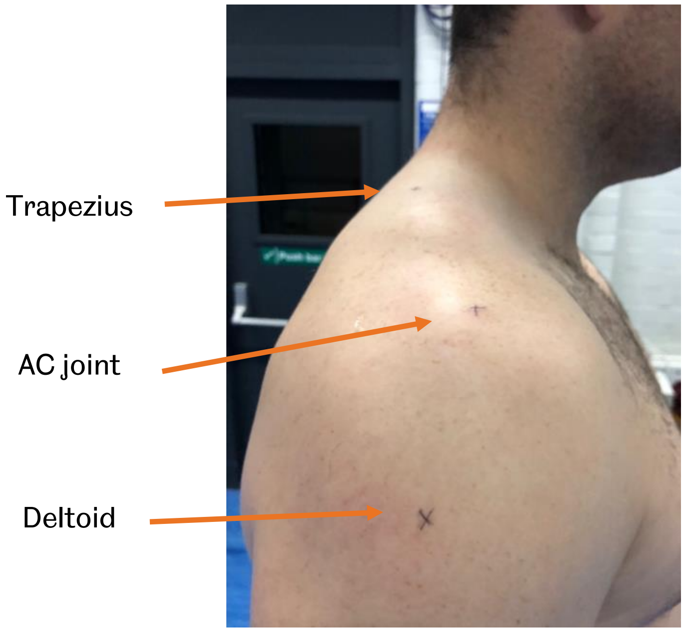


Figure 4.8 – Anatomical landmarks used for scans.

Ultrasound gel was applied to the transducer and placed with as little pressure as possible to not distort the internal geometries of each landmark, ultrasound images could then be captured. Full training on the ultrasound system was given before testing. Images were then exported to an image processing software and measurements of tissue depths were made. Figure 4.9 displays an ultrasound image taken superior to the central body of the trapezius muscle, from which layer thickness can be calculated due to the overall depth of the ultrasound image being known.

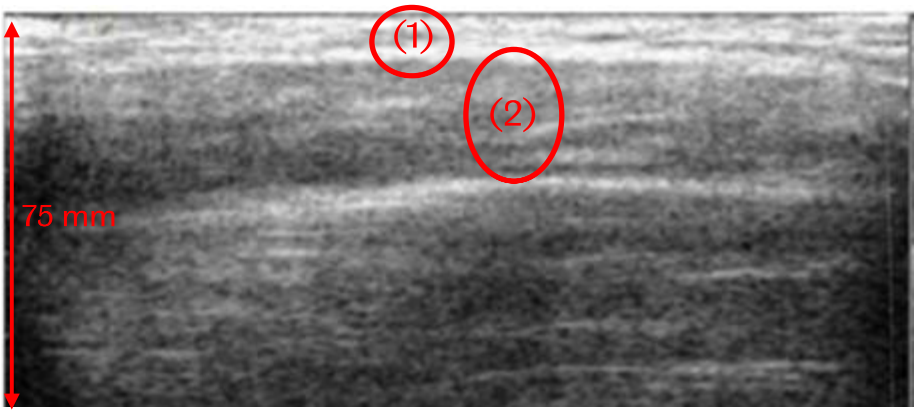


Figure 4.9 - An ultrasound image of tissue layers taken superior to the central body of the trapezius muscle layer, (1) - Skin, Adipose and Fascia Layer, (2) - Muscle Layer. *(Note: bone is not visible).*

Results

Table 4.4 provides the mean layer thicknesses found in the ultrasound scanning of shoulders. The superior element of the shoulder contains the Trapezius and AC joint while the lateral element contains the Deltoid as highlighted in Figure 4.6. There is no muscle in the AC joint so muscle layer thickness is zero.

Table 4.4 - Shoulder layer thickness measurements, from ultrasound scans (n=6). Mean ± Standard Deviation (SD).

	Trapezius region		**Deltoid region**	
	Muscle	**Skin, adipose tissue and fascia**	**Muscle**	**Skin, adipose tissue and fascia**
Thickness (mm)	17.0 ± 2.0	5.6 ± 2.0	33.2 ± 2.7	6.0 ± 1.7

Key average layer thicknesses presented in the results section give key information for developing a shoulder surrogate. The shoulder is however very complex with layer thickness varying throughout and from person to person. Taking into account the results from external body scanning, simplifications must be made to develop a repeatable shoulder surrogate that can be made easily. Therefore, these tissue thicknesses guided the final design of a simplified shoulder surrogate (§5.4).

The data gathered can also be used to compare against the general population. Average Trapezius and Deltoid muscle tissue thicknesses have been found to be 11.9 mm [191] and 29.0 mm [192] in the general population. This is 5.1 mm and 4.2 mm less than what was found in the current research.

Intra and Inter Observer Reliability

Intra-observer and Inter-observer reliability were both assessed using measurements from the muscle tissue depth of each participant on their deltoid. Cohen's Kappa Coefficient (K) was used as a measure of reliability (equation 4.1)

$$K = \frac{Po - Pc}{1 - Pc} \tag{4.1}$$

Where P_o is the percentage value of agreement and P_c is the percentage value of expected agreement by guessing. For Intra-observer reliability, measurements of six participants' deltoid muscle depth were re-analysed by the primary researcher. Classification of Kappa values [193] showed a very good level of agreement (K = 0.95). Inter-observer reliability was assessed by a secondary researcher. The secondary researcher also had training on the ultrasound equipment and a very good knowledge of human anatomy. Again, a very good level of agreement was found (K= 0.88).

4.5 Mechanical Characterisation of Organic Tissues

4.5.1 Overview

Organic tissue properties of muscle, adipose, skin, and bone (cortical) have been measured through obtaining new data as well as adapting previous research. Given the large degree of variation in tissue properties between many characterisation studies, it is clear there is no consensus on their definitive mechanical properties. Therefore, in this study, an absolute representative data set has been selected for each type of tissue. Where necessary, new data has been obtained by the author. The mechanical properties of organic tissues were attained using uniaxial compression tests.

Skin and muscle have been reported to exhibit anisotropic behaviour, whilst adipose tissue has been reported as isotropic when subjected to loading. Muscle and skin tissue have been reported to have low compressibility [110, 117, 194], therefore having a Poisson's ratio estimated at 0.5. Adipose tissue has been reported to be more compressible, having a lower Poisson's ratio estimated at 0.38 [195]. Adipose tissue generally undergoes non-recoverable deformations [196].

Compressive engineering stress-strain data for muscle tissue has been obtained by the author, while stress-strain data from Shergold et al. [197] Comley & Fleck [133], and McElhaney [147] has been collated to represent bone, skin and adipose tissue. The resulting dataset has then been used to guide the development of shoulder surrogate simulant materials in chapter 5.

4.5.2 Muscle

Compressive data from relaxed ex vivo Porcine muscle tissue was obtained. Muscle contraction was not considered due to both the lack of data and difficulty in obtaining this data. This also causes an issue with variability in the data due to the level of contraction not being known. As well as this, it has been suggested that the human monosynaptic stretch reflexes occur between 30 - 60 ms, therefore in most sports impacts, the most acute compressive stage will occur prior to the body responding to the impact itself [198]. The methods and results are explained below.

Deceased organic porcine shoulder muscle tissue was taken from a 2-year-old pig slaughtered 2 days before testing, during this time is was stored in a fridge at 4°C. The porcine muscle tissue was cut using bespoke stamps to make cylindrical test specimens (29 mm Ø, height of 12.5 mm) for quasi-static testing and cubic test specimens (width of 7 mm, length of 7 mm, height of 12.5 mm) for dynamic mechanical analysis. All test specimens were measured using digital callipers (Mitutoyo, Takatsu-ku, Japan) and cut from the same region of the porcine shoulder. All the samples were acclimtised to room temperature prior to testing.

A Shimadzu test machine (Shimadzu, EZ-LX, Kyoto, Japan, 1 kN load cell) (Figure 4.10a) was used to measure the compressive response of the porcine samples. A compressive test protocol was used, increasing the engineering strain until material failure was achieved. This was repeated for 5 different test specimens at 5, 20, 50 mm/min, and a median result was taken. Force, displacement, and time outputs were taken from the test machine to calculate engineering stress and strain (Figure 4.10b). Test specimens were placed in a Dynamic Mechanical Analyser (Metravib, VA2000, Limonest, France) to characterise the compressive response at a dynamic strain rate. To observe the effects of dynamic strain on the dynamic mechanical properties of the specimens, a strain sweep from 0% to 1% was performed at 10 Hz at room temperature. Results are seen in Figure 4.11.

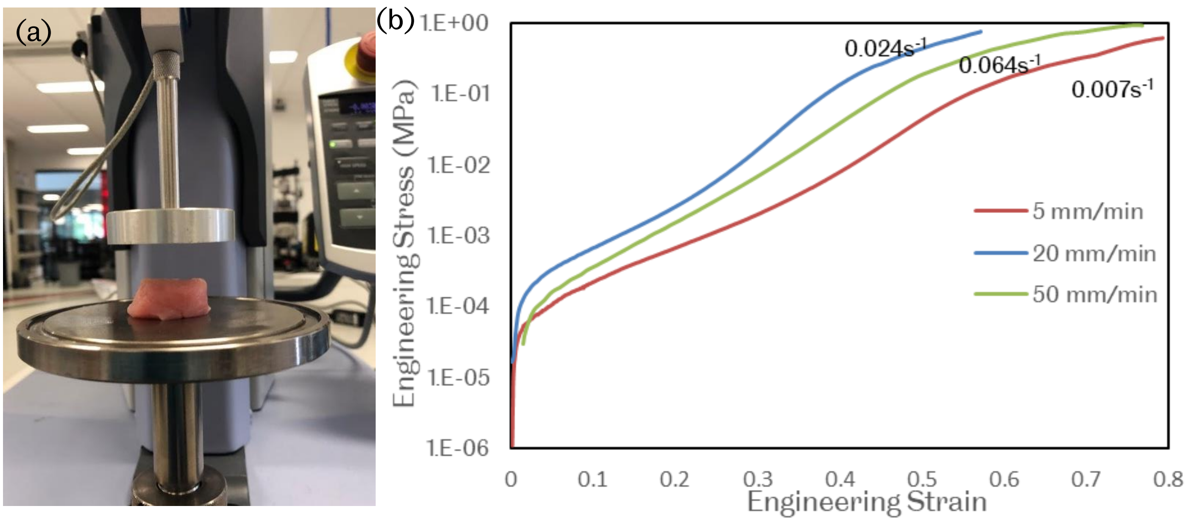


Figure 4.10 – (a) Uniaxial Compressive test setup (b) Log-linear compressive stress-strain plot of porcine muscle at varying strain rates.

The compressive response of porcine muscle was found to be strain rate dependent but, this did not seem to follow a continuous trend. This is highlighted by the data provided in Table 4.5 where the compressive modulus at 50 % strain increases from 0.09 MPa to 0.86 MPa from 5 mm/min to 20 mm/min test speed but then decreases to 0.37 MPa at 50 mm/min. This demonstrates the variability that can be seen in organic tissues. It is important to note this strain rate is far slower than in a typical impact in rugby where average tackle velocities will range between 4.8 (ball carrier) – 5.2 (tackler) m/s, potentially causing strain rates in soft tissues between 288000 – 336000 mm/min [56]. This making it important to also test organic tissue at more dynamic strain rates.

Table 4.5 – Compressive Modulus of Porcine Muscle at 50 % Strain.

	Strain Rate (mm/min)		
	5	20	50
Compressive Modulus (MPa)	0.09	0.86	0.37

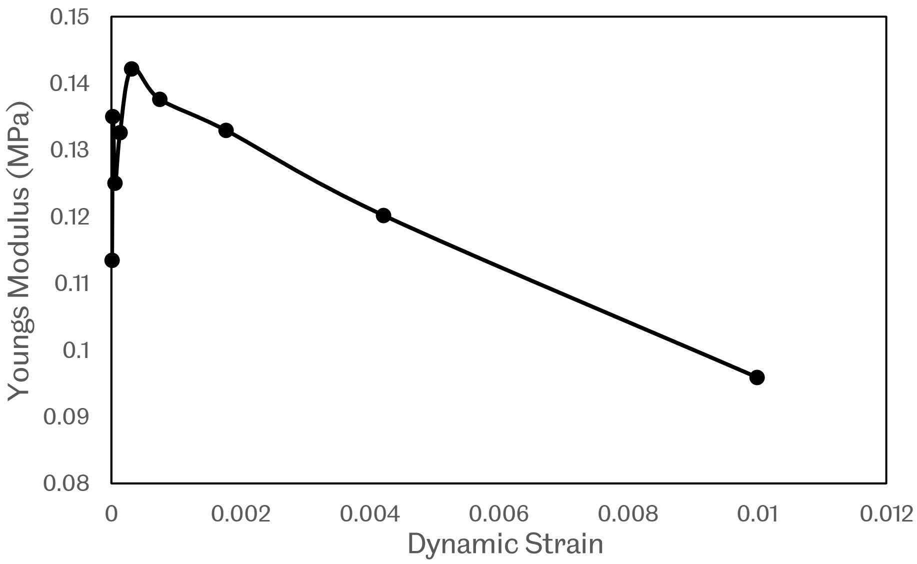


Figure 4.11 – Compressive Young's Modulus strain sweep of porcine muscle tissue.

4.5.3 Adipose Tissue

The compressive stress-strain response of organic adipose tissue (porcine) is detailed in Figure 4.12. The Figure has been adapted from Comley & Fleck [133] and data from intermediate strain rates used on a logarithmic scale.

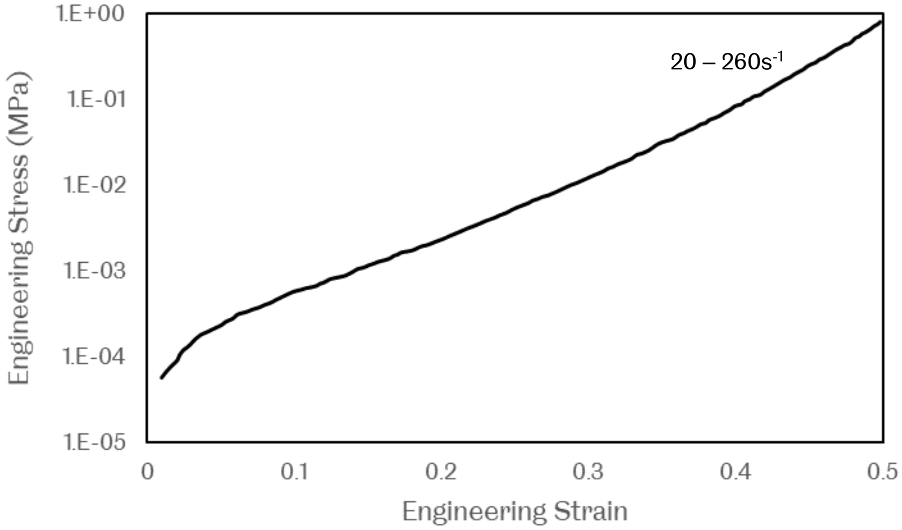


Figure 4.12 – Log-linear compressive stress-strain plot of porcine adipose tissue at intermediate strain rates [133].

4.5.4 Skin

The compressive stress-strain response of organic skin (porcine) is detailed in Figure 4.13. Porcine abdomen tissue was used for this testing, it has a similar thickness and mechanical properties to its human counterpart [199]. The Figure has been adapted from Shergold et al. [197] and data from intermediate strain rates used on a logarithmic scale.

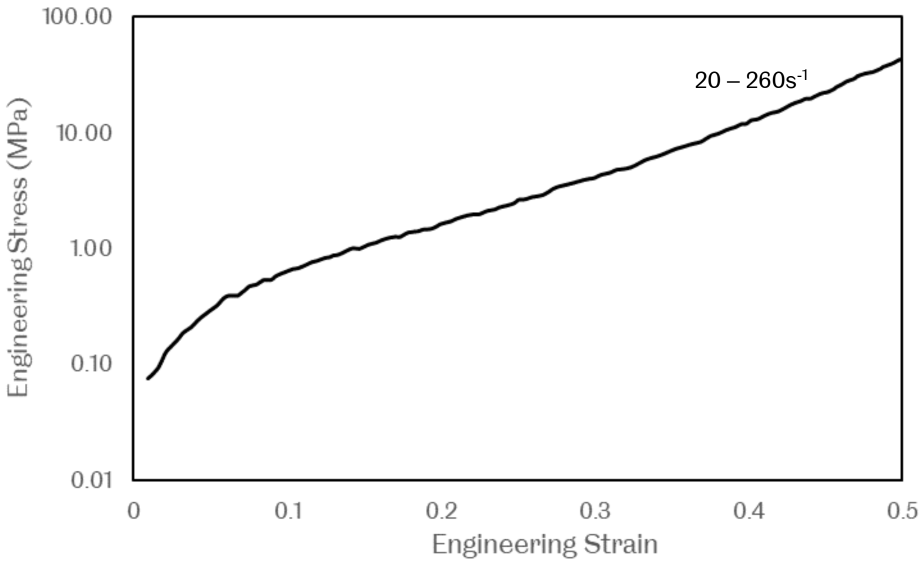


Figure 4.13 – Log-linear compressive stress-strain plot of porcine skin tissue at intermediate strain rates [197].

4.5.5 Bone

The compressive stress-strain response of organic cortical bone (human) is detailed in Figure 4.14. The Figure has been adapted from McElhaney [147] and data from quasi-static and intermediate strain rates used on a linear scale. Cortical bone is significantly stiffer than trabecular bone. Due to this, it is believed it contributes to a significant part of the tissue's response to load [200]. In this case, its compressive response was therefore detailed.

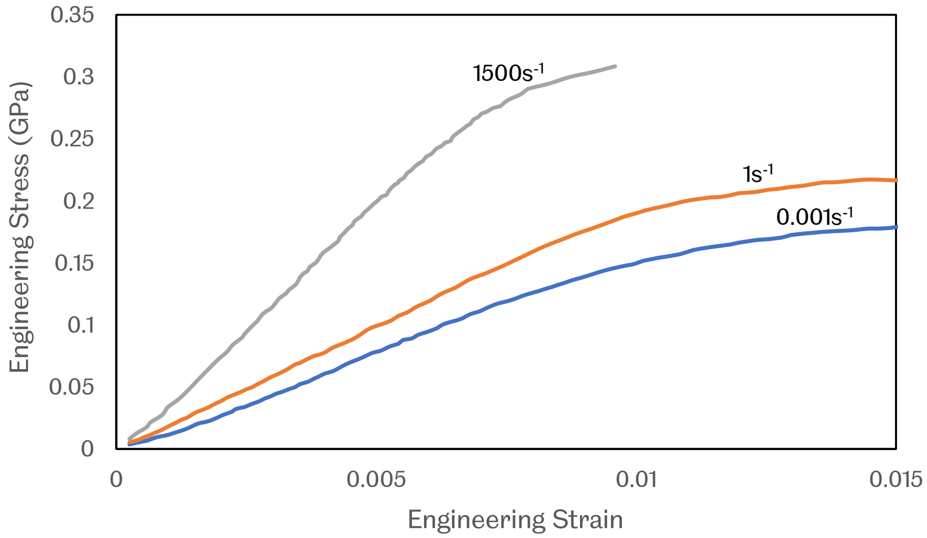


Figure 4.14 – Compressive stress-strain plot of human cortical bone at differing strain rates [147].

4.5.6 Target Dataset

Key data regarding organic tissues mechanical response is listed in Table 4.6. Due to the strain-dependent nature of organic tissues, a range in values has been selected. When developing simulants that act as an impact surrogate, it is important that the compressive response of the simulant closely matches that of the organic tissue it embodies. Obtaining compressive stress-strain data was therefore paramount to the study.

Table 4.6 – Key mechanical properties of organic tissues.

Organic Tissue	Property	Value
Muscle (relaxed)	*Density (g/cm^3)*	1.06 [130]
	Compressive Modulus (MPa)	0.09-0.86*
	Tensile Strength (MPa)	0.44 [201]
Adipose Tissue	*Density (g/cm^3)*	0.93 [202]
	Compressive Modulus (MPa)	0.002-4 [133]
Skin	*Density (g/cm^3)*	1.02 [112]
	Compressive Modulus (MPa)	0.0083 [203]
	Tensile Strength (MPa)	18-36 [119]
Bone (cortical)	*Density* (g/cm^3)	1.47-2.12 [142, 143, 204]
	Compressive Modulus (MPa)	11.5-17 [143, 205]
	Compressive Strength (MPa)	100-182 [206]
	Impact Toughness Kc (MPa $m^{1/2}$)	2.16-9.0 [207]
	Shore D Hardness	85-95 [206]
	Tensile Strength (MPa)	88-151 [131, 146]
	Tensile Modulus (GPa)	18.6 – 20.7 [208]

**Data taken from new data obtained by the author.*

The mechanical properties of organic tissues depend on many variables. It is clear that both from a review of the literature and the testing of porcine muscle tissue, the strain rate of testing can vary organic tissue's response to load. When developing simulant materials for impact testing, the nature of the impact and its strain rate needs to be considered.

The data collected on the mechanical properties of organic tissues in this chapter was used to guide the development of organic tissue simulant materials leading to the fabrication of human shoulder surrogates (§5). Obtaining this data was therefore a key part of this process.

While not included in the data presented, factors such as muscle contraction or the mechanical response of layer of organic tissue in a system could also be considered when collecting data to develop simulant materials to be used as a human impact surrogate. This should be considered in future studies.

4.6 Summary

This chapter provides an assessment of both the human shoulder and the mechanical properties of organic tissues to develop a shoulder surrogate for the evaluation of padded clothing in an impact testing setup. The following presents an evaluation of how the testing completed will help to develop a shoulder surrogate as well as the limitations that may occur.

4.6.1 External Assessments of the Shoulder

Benefits

- The 3D scans taken of rugby players' shoulders can be easily 'reversed engineered' to develop an accurate mould for a shoulder surrogate.
- The external geometries collected from scanning can be used to guide the simplification of geometries in a shoulder surrogate to make fabrication procedures easier.

Limitations

- The scanning was only completed on nine participants. A larger sample size would allow for the geometrical comparison of positional and BMI differences. A comparison could also be made to the general public.

4.6.2 Internal Assessments of the Shoulder

Benefits

- The ultrasound scans taken of rugby players' shoulders provide insight into internal tissue layer thicknesses and can help guide shoulder surrogate development.

Limitations

- Ultrasound scans only provide a guide on tissue layer thicknesses and do not create a 3D assessment of the internal geometries in rugby players.

- The scanning was only completed on six participants. A larger sample size would allow for the geometrical comparison of positional and BMI differences. A comparison more in-depth could also be made to the general public.

4.6.3 Mechanical Assessments of Organic Tissues.

Benefits

- The testing completed provides an understanding of the compressive behaviour of organic (porcine) muscle tissue at differing strain rates. This can be used to help develop organic muscle tissue simulants with similar compressive properties. This is key when fabricating an impact surrogate.

Limitations

- Mechanical testing at non-harmful loads was not completed on human shoulders due to the Covid-19 pandemic. This data could be used to validate the shoulder surrogates developed in this thesis. This could have been performed by using a simple Shore A durometer on human participant's skin to get hardness properties or by developing a force-displacement rig that could be used on live human subject's shoulder area.
- Mechanical testing was only completed on organic muscle tissue. Data from other studies were taken for other organic tissues and full testing protocols were not known.

CHAPTER 5 - HUMAN SHOULDER SURROGATE FABRICATION AND FORMULATION

5.1. Chapter Overview

This chapter presents the development of custom-made silicones tailored to match the response of human soft tissue. This process is documented thoroughly explaining the iterative methods and mechanical testing completed to achieve the final silicones. The chapter highlights the considerations taken when other soft tissue simulants were used in the surrogate. The chapter then details the design and validation processes of both a simplified shoulder surrogate and a biofidelic shoulder surrogate for use in impact testing of padded rugby clothing.

5.2 Surrogate Soft Tissue Formulation

5.2.1 Introduction

The simulant materials used are a key component in developing a biofidelic surrogate. The choice of material can significantly affect the surrogate's mechanical behaviour. The simulant material must match the dynamic stiffness of the tissue it represents to provide a biofidelic response to impact [106]. Many studies have selected a silicone using either static or linear stiffness parameters [104, 209]. However, as silicones are non-linear elastomers the correct mechanical properties may not be matched. Ensuring the simulant material matches this non-linear response to load provides a key requirement for a human surrogate.

The field of quantifying the mechanical behaviour of human soft tissues is ever-growing, with a large amount of variation between many characterisation studies. As well as this, factors like soft tissue tonicity and their anisotropic behaviours arguably make developing simulants with exact mechanical properties to the soft tissue they are embodying some years away. Taking this into account the current research aims to develop soft tissue simulants that represent the mechanical properties established in §4.4.

5.2.2 Polysiloxane

Polysiloxane or silicone are polymers made up of Siloxane (−R2Si−O−SiR2−, where R = organic group). Silicones consist of an inorganic silicon–oxygen backbone chain (⋯−Si−O−Si−O−Si−O−⋯) with two organic groups attached to each silicon centre.

Commonly, the organic groups are methyl. The materials can be cyclic or polymeric. Through the altering of constitutive silicone components, it is possible to modify the mechanical properties of the resulting elastomer so that they match the human counterpart they represent.

Silastic™ 3481 (Dow Corning, UK) is an 'off the shelf' silicone that has previously been used in the sports industry to represent human soft tissue is a two-part additive cure silicone, it cures at room temperature by condensation reaction when a curing agent is added to its base. Additive cure silicones provide a good solution for use as human tissue surrogates as the fabrication process is simple, there is no shrinkage so accurate geometries can be achieved and their compressive properties are to some extent similar. Silastic™ 3481's chemical structure is proprietary. However, it is common knowledge (stated in the MSDS) that it contains Dodecamethylcyclohexasiloxane ($C_{12}H_{36}O_6Si_6$), Octamethylcyclotetrasiloxane $[(CH_3)_2SiO]_4$, and Decamethylcyclopentasiloxane $[(CH_3)_2SiO]_5$. All three of these are silicones compounds. Their chemical structures are in Figure 5.1.

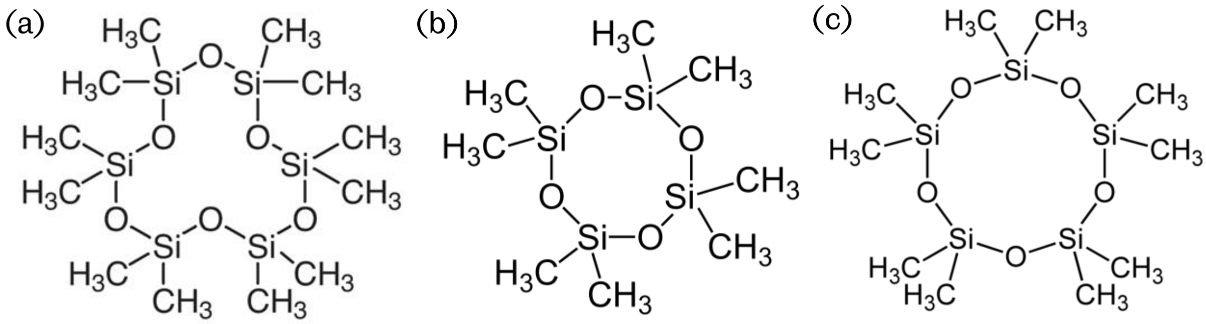


Figure 5.1 – Chemical structure of (a) – Dodecamethylcyclohexasiloxane, (b) – Octamethylcyclotetrasiloxane, (c) – Decamethylcyclopentasiloxane.

Fourier-transform infrared spectroscopy (FTIR)

Samples of Silastic™ 3481 were cured via the use of a curing agent resulting in a crosslinked Polysiloxane. These samples were studied using FTIR to identify the compounds found in the elastomer. The resulting spectra are shown in Figure 5.2. From these spectra, characteristic peaks are seen at 2962, 1258, 1010, and 787 cm^{-1}. Table 5.1 highlights what compounds these peaks suggest are present in Silastic™ 3481.

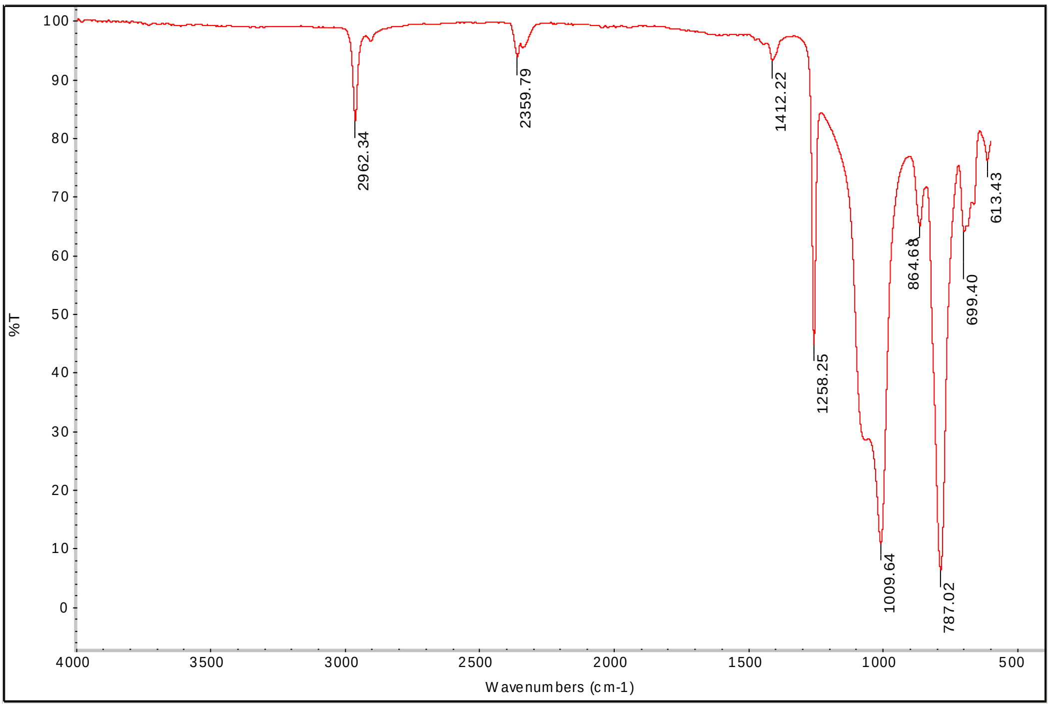


Figure 5.2 – FTIR spectra of Silastic™ 3481.

Table 5.1 – FTIR peaks identified by wavenumber peak location for Silastic™ 3481.

Peak	**Peak Location (cm^{-1})**
CH_3 stretching	2962
CH_3 deformation in Si-CH_3	1258
Si-O-Si stretching	1010
-CH_3 rocking and Si-C stretching in Si-CH_3	787

To alter the mechanical properties of two-part additive cure silicones, deadener can be added to its constituents. Deadener will soften the silicone and reduce its stiffness. Deadener does this by inhibiting the cross-linking of polymer chains when curing occurs. The greater the ratio added, the greater the effect on the silicone's properties. Hardener can also be added to the silicones constituents to stiffen the resulting elastomer.

5.2.3 Silastic™ 3481 Fabrication Procedures

Compressive test samples were fabricated to provide a direct comparison to the previous testing completed in §4.5. Due to previous studies finding that Silastic™ 3481 had stiffer properties to relaxed muscle tissue, deadener was added in varying quantities. This created a three-part blend of base, catalyst, and deadener. These blends were set into ASTM D395 (29mm Ø, 12.5mm height) cylindrical test moulds as prescribed for compressive testing of rubber compounds (Figure 5.3). For tensile testing, these blends were set in dog bone shaped moulds (10 mm width, 120 mm length) and for dynamic mechanical analysis they were set in cubic test moulds of 7 mm width, 7 mm length, and 12.5 mm height.

There are many different elements of this process that may affect the mechanical properties of the cured silicone. These elements may be, but are not limited to temperature, curing time, weighing inaccuracies, mixing methods, and degassing processes. The steps taken to cure the silicones are set out below:

- Components (base, catalyst, deadener) are weighed (± 0.1 g accuracy) and poured into a container.
- Components are mixed manually for 5 minutes until the mixture is consistent.
- Silicone mix is fully degassed using a vacuum degassing chamber (DVP, EC20, Stoke-on-Trent, UK).
- Moulds are coated with silicone release spray (Rocol, Mould Release Agent, UK).
- Silicone mix is poured into mould straight after degassing is complete.
- Silicone mix is left in moulds at room temperature (23°C ± 2°C) for 24 hours to cure.
- The cured specimen is removed from moulds, cleaned, and left for a further 7 days to fully cure.

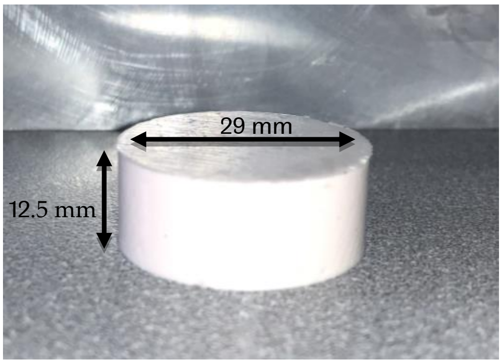


Figure 5.3 - ASTM D395 Silicone test sample

5.2.4 Mechanical Testing Procedures

`a) Quasi-Static Compression Tests

Quasi-static compression tests were conducted at varying strain rates (5 mm/min – 50 mm/min) using a Shimadzu test machine (Shimadzu, EZ-LX, Kyoto, Japan, 1 kN load cell) (Figure 5.4). A range of load cells was used (0.5 kN – 5 kN) dependent on the test sample. Each sample was placed between flat compressive platens before being coated in Vaseline to reduce friction. The software was programmed to perform a compressive test protocol until 0.8 strain was achieved. This procedure was used to investigate the silicone's response to compressive load at differing quasi-static strain rates.

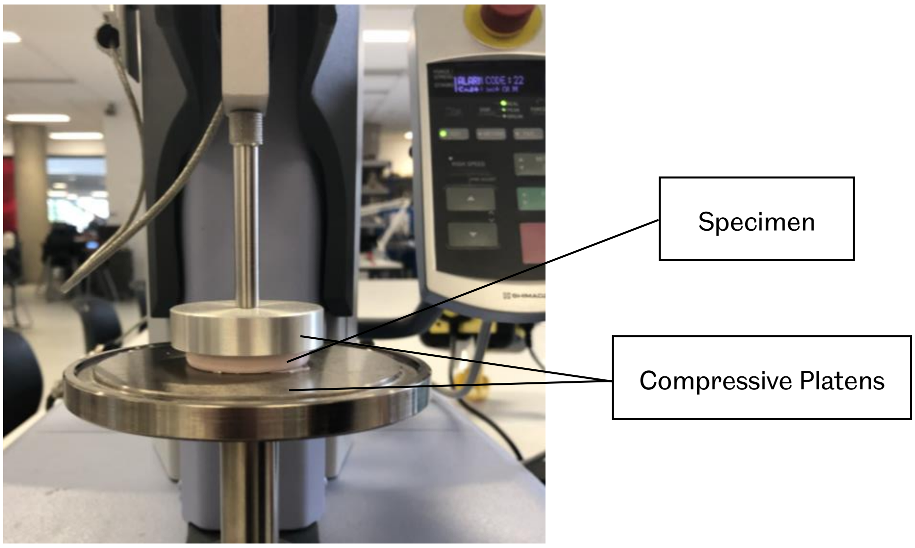


Figure 5.4 – Shimadzu test set up with compressive test discs.

b) Quasi-Static Tensile Tests

Quasi-static tensile tests were conducted at varying test speeds (5, 20, 50 mm/min) using a universal testing machine (Hounsfield, H10Ks, USA, 5 kN load cell). Each sample was loaded into the test machine and the initial gauge length measured. The specimen was held in by self-tightening grips, with the upper one connected to a load cell. Each test was performed until the specimen broke. The procedure was used to investigate the silicone's elastic tensile properties at quasi-static strain rates.

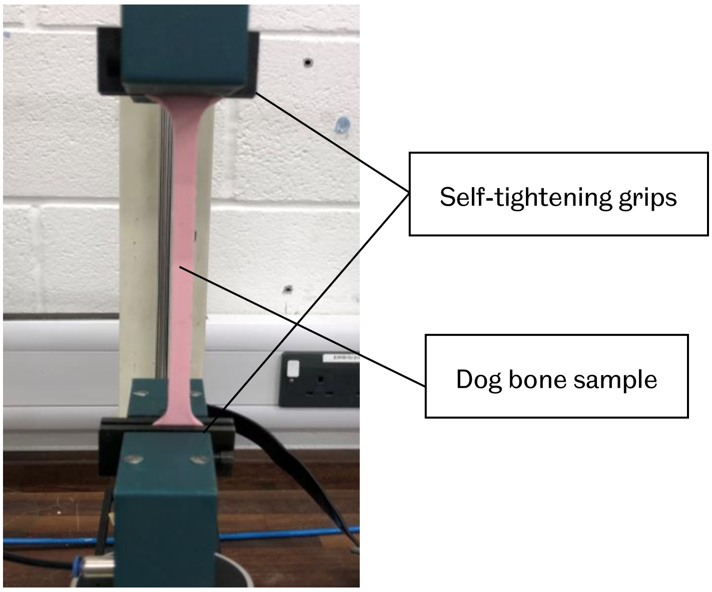


Figure 5.5 – Mechanical test set up with tensile dog bone samples.

c) Stress Relaxation Tests

Stress relaxation tests are used to determine the viscoelastic properties of a material. Uniaxial ramp-and-hold stress relaxation tests were performed on ASTM D395 silicone test samples to understand the viscoelastic time-dependent properties of the silicones. Each silicone specimen was deformed to 0.5 strain at a rate of 1000 mm/s and held at a constant strain for 50 seconds. This was deemed sufficient time for the silicone to achieve a steady relaxed state [194].

d) Dynamic Mechanical Analysis

Moulded test cubes of silicone test moulds in a Dynamic Mechanical Analyser (Metravib, VA2000, Limonest, France) to characterise the compressive response at a dynamic strain rate. To observe the effects of dynamic strain on the dynamic mechanical properties of the specimens, a strain sweep from 0% to 1% was performed at 10 Hz at room temperature. This is due to impact testing with the addition of the shoulder surrogates being performed at room temperature.

e) Analysis Procedures

Force – displacement data were obtained from the aforementioned tests. Both the engineering strain (ε) and engineering stress (σ) were calculated at each time interval. The engineering strain (Equation 5.1) was calculated using the displacement values taken from the machine and the original length measured using digital Vernier callipers ($\pm 1 \times 10^{-5}$ m).

Engineering stress was calculated (Equation 5.2) using load cell data and the cross-sectional area (mm^3) of the sample.

$$\varepsilon = \frac{\Delta \iota}{\iota_0} \tag{5.1}$$

Where: l_0 = original length; Δl = change in length

$$\sigma = \frac{\mathrm{F}}{\mathrm{A}_0} \tag{5.2}$$

Where F = applied load; A_0 = original cross-sectional area

5.2.5 Silicone Formulations

Base (Silastic™ 3481, Dow Corning, UK), Catalyst (RTC 10 Curing Agent. Dow Corning, UK), Deadener (Platsil Gel 25 LV, Mouldlife, Suffolk), and Hardener (Platsil Gel 25 Part H, Mouldlife, Suffolk) were used in varying concentrations to formulate silicones of differing mechanical properties. All the formulations constituents were measured and defined using a weight ratio. Each silicone had a standard mix ratio of 10:1 (Base : Catalyst), with the addition of deadener or hardener making up the third constituent. I.e., 10:1:4D would be ten parts Base, one-part Catalyst, and four parts Deadener. Each silicone's constituents are outlined in Table 5.2. To calculate the density, Three ASTM D395 silicone test samples were formulated, and dimensions were measured using callipers. They were then weighed using an analytical balance (Fisherbrand, Analytical series, UK, accuracy ± 0.0001). The mean is shown in Table 5.2.

Table 5.2 – Silicone sample constituents as a weight ratio and density.

Sample number	**Constituents**	**Density ± SD (n = 3) (kg.m^3)**
1	10:1	1120 ± 18
2	10:1:2D	1068 ± 30
3	10:1:4D	1072 ± 75
4	10:1:8D	1054 ± 23
5	10:1:2H	1222 ± 14

**D = Deadener, H = Hardener*

5.3 Mechanical Properties of Silicones

5.3.1 Quasi-Static Compression Tests

Engineering stress-strain graphs for quasi-static uniaxial compression tests on each silicone at test speeds of 5, 20, and 50 mm/min have been plotted for each silicone formulation (Figure 5.6). The graphs illustrate that the addition of deadener in increasing ratios to the base and catalyst decrease the stiffness of the final sample. This decrease in stiffness is displayed throughout all the strain. It should be noted the addition of a hardener did not increase the stiffness of the sample and actually reduced it.

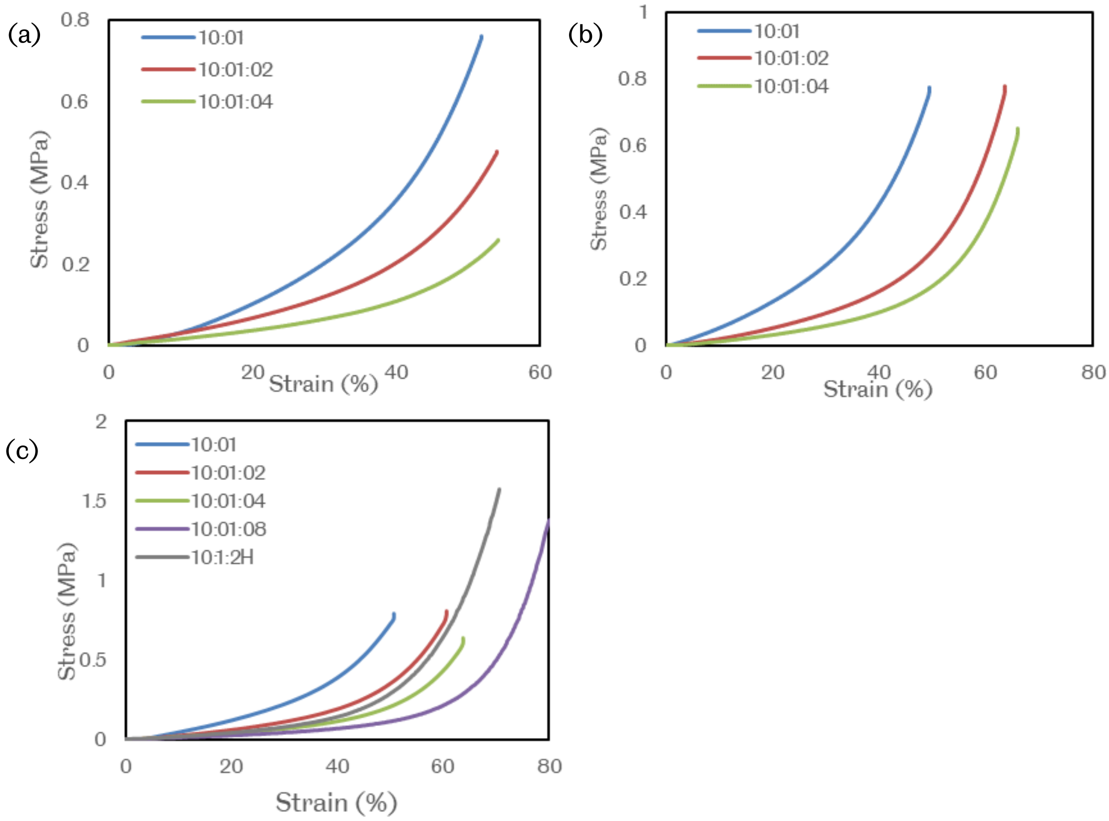


Figure 5.6 Quasi-static compression plots of silicones at (a) 5mm/min test speed, (b) 20mm/min test speed, (c) 50mm/min test speed.

When both formulations of 10:1 and 10:1:4 are plotted at differing quasi-static test speeds (Figure 5.7) similar plots are seen. This suggests there is no strain rate effect at quasi-static test speeds, over the range tested.

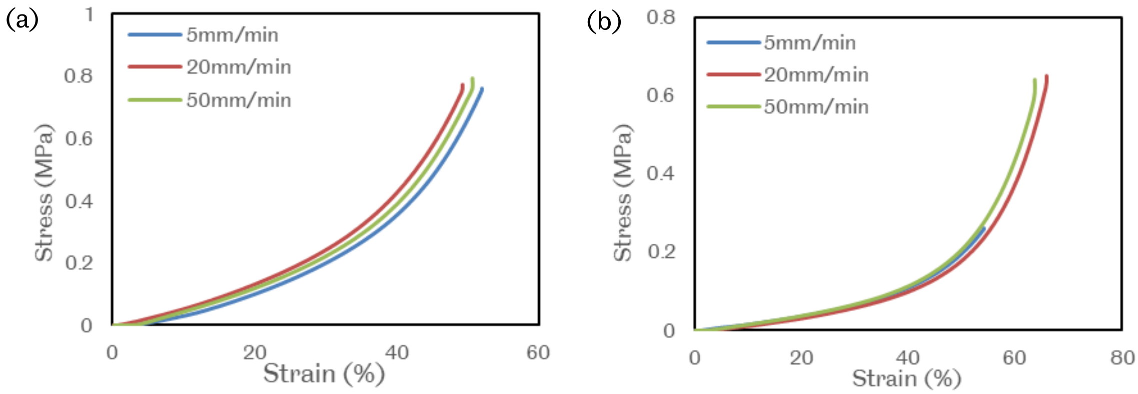


Figure 5.7 Quasi-static compression plots at varying test speeds of (a) Silastic™ 3481 (10:1) and (b) Silastic™ 3481 + Deadender (10:1:4).

Table 5.3 – Compressive modulus of each silicone formulation at 0.5 strain

Formulation	**Compressive Modulus* (MPa)**
10:1	0.97
10:1:2D	0.48
10:1:4D	0.28
10:1:8D	0.17
10:1:2H	0.35

**Test speed of 50mm/min*

5.3.2 Quasi-Static Tensile Tests

Stress-strain graphs for quasi-static uniaxial tensile tests on each silicone at a test speed of 50 mm/min have been plotted (Figure 5.8a), and at varying test speeds for 10:1:4. The tests were performed until the failure of the test sample. There is a clear reduction of stiffness when deadener concentration is increased. Failure also occurs at smaller strains when increasing the deadener concentrations suggesting it lowers its UTS, this is also shown in Table 5.2, Although this is not the case at the higher levels of deadener concentrations between samples 10:1:4 and 10:1:8. As with the compression testing completed (§5.3.1), the addition of the hardener decreased the samples' stiffness.

There is no clear difference in the plots when comparing different quasi-static test speeds (Figure 5.8b). The Young's Modulus values are shown in Table 5.3 at an engineering strain of 1. Increasing the deadener concentration lowers the stiffness of the sample.

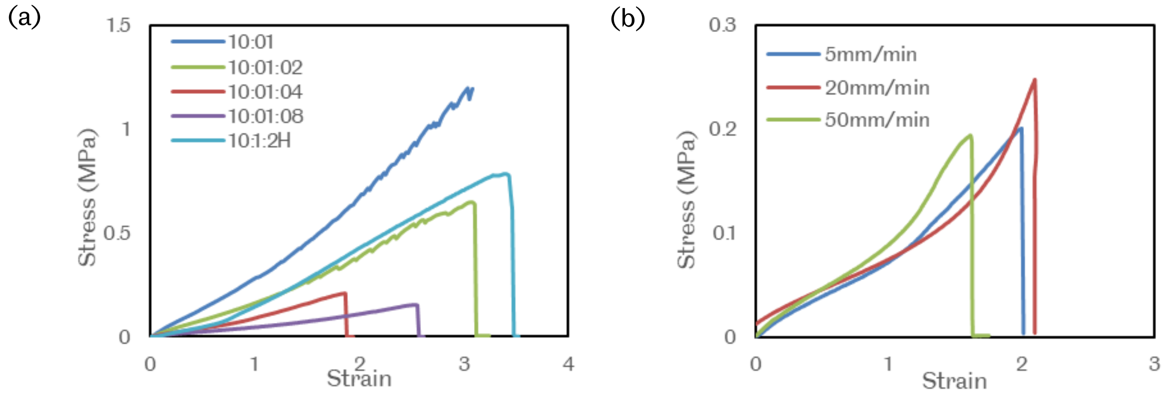


Figure 5.8 - Quasi-static tensile plots of (a) Silastic™ 3481 formulations at 50mm/min test speed and (b) Silastic™ 3481 (10:1:4) at varying test speeds.

Table 5.4 – Young's modulus and UTS values for each silicone formulation.

Formulation	Young's Modulus (MPa)	Ultimate Tensile Strength (MPa)
10:1	0.284	1.196
10:1:2D	0.163	0.647
10:1:4D	0.089	0.194
10:1:8D	0.048	0.155
10:1:2H	0.161	0.631

5.3.3 Stress Relaxation Tests

Stress relaxation tests on 10:1:4 formulation (solid line) specimens are compared with relaxed porcine muscle tissue (dashed line) (Figure 5.9). This porcine tissue data has been taken from previous research by Van Loocke et al. [194]. The silicone formulation exhibits a smaller engineering stress; however, a much shorter stress decay is observed. Stress decays appear to stop occurring after 6 seconds. Stress decay is still present in organic tissues past 60 seconds. This is possibly due to the in-vitro nature of organic tissues and the re-distribution and/or expulsion of fluid inside the tissue sample during a stress relaxation test.

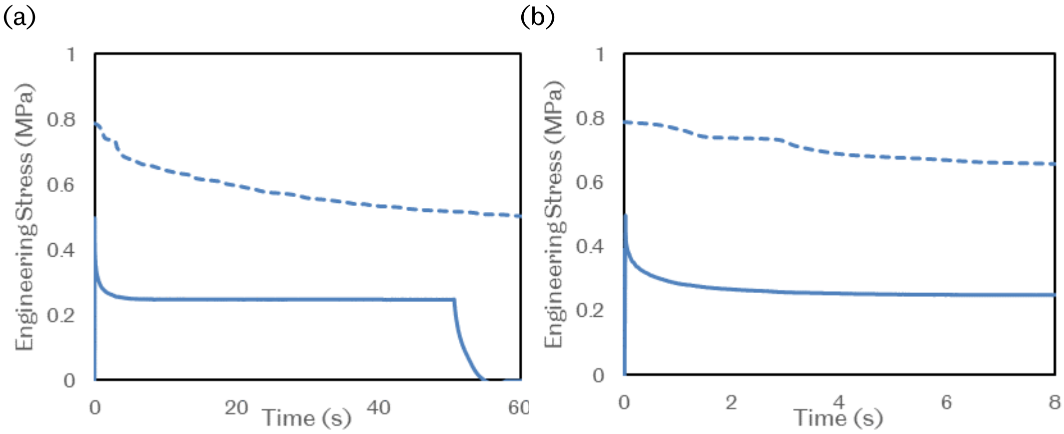


Figure 5.9 – Engineering stress-time plot showing stress relaxation of formulation 10:1:4 *(Solid line)* and relaxed porcine muscle tissue [194] *(Dashed line)* for (a) 60 seconds and (b) 8 seconds.

5.3.4 Dynamic Mechanical Analysis

Figure 5.10 displays the dynamic strain against the compressive Young's Modulus of formulations 10:1 and 10:1:4. The results show a clear significant difference between the two formulations, A larger Young's Modulus is seen in the 10:1 sample throughout the strain sweep suggesting an increase in stiffness.

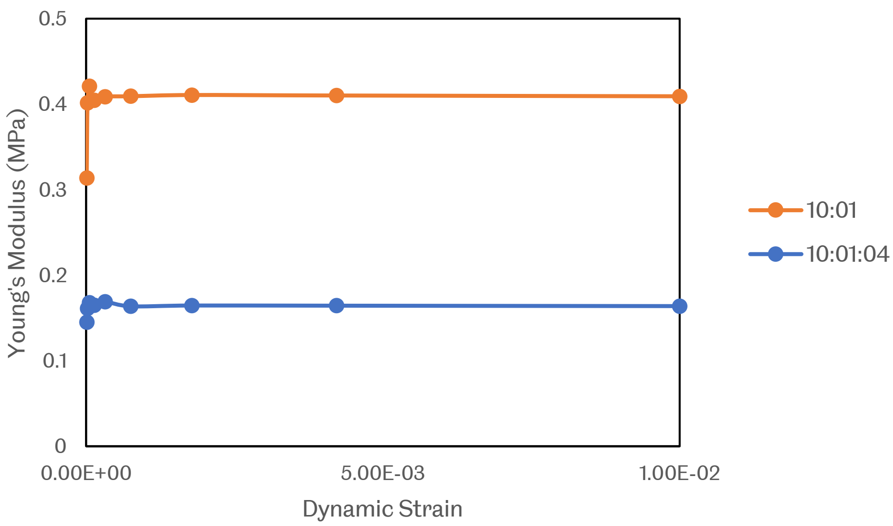


Figure 5.10 - Strain sweep of Young's Modulus comparing Silastic™ 3481 (10:1) and Silastic™ 3481 + Deadener (10:1:4).

5.3.5 Bespoke Silicone for Organic Tissues

After mechanical testing was completed on each silicone formulation, the results were compared with organic tissue data to establish which silicone formulations should be used as an organic tissue surrogate. The target data set was to fit the relaxed muscle dataset established in §4.5.6. When comparing results from organic tissue data the silicone formulation 10:1:4 had the closest fit both at quasi-static and dynamic strain rate tests. The following Figures (5.11 – 5.13) display this with absolute error (%) overlayed. The quasi-static compressive response of Porcine muscle tissue and formulation 10:1:4 was different at low strains, with Porcine tissue being less stiff, this perhaps due to its lack of muscle tonicity as it is ex-vivo. However, at strains of 0.5 and over a much closer match is shown. The silicones application as part of an impact surrogate must be considered here. Organic tissues have demonstrated large strain deformations when subject to impacts [128]. High deformations and therefore strains (> 50 %) are seen in many sports impacts [159] so this difference at low strain is less of a concern.

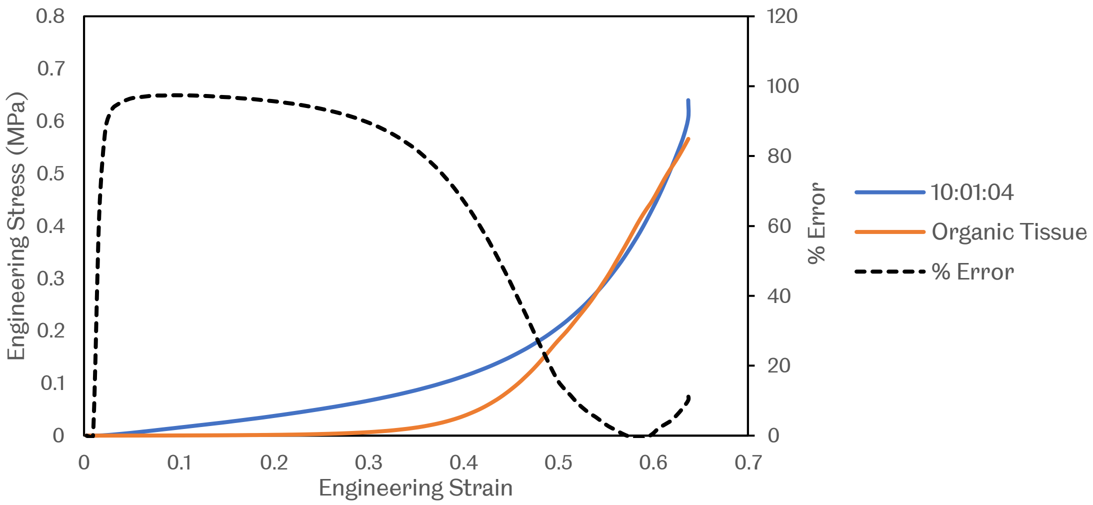


Figure 5.11 – Quasi-static stress-strain compression plot (50 mm/min) of relaxed porcine muscle tissue, 10:1:4 silicone formulation with absolute error (%) overlayed.

There is a substantial error and therefore differences in the tensile properties of organic tissue data and 10:1:4 at low strains. Although these results differ the resultant use of the silicone simulant in an impact surrogate where compressive loading will take place must be considered. It is therefore important the compressive properties of the silicone are similar to organic tissue and altering their tensile properties may change this. It is also important to note that the maximum strain shown in the graph is 0.25 due to the availability of literature.

Organic tissues tensile response at higher strains are therefore unknown.

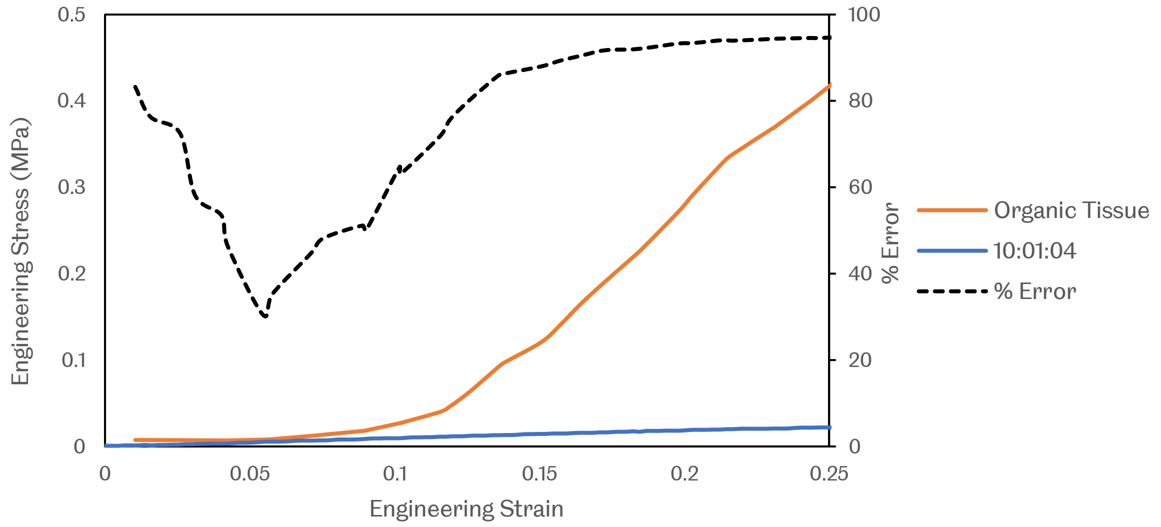


Figure 5.12 – Quasi-static stress-strain tensile plot of relaxed porcine muscle tissue, 10:1:4 silicone formulation with % error overlayed.

Dynamic mechanical analysis shows organic tissue has a slightly lower compressive Young's Modulus at dynamic strains than the 10:1:4 silicone formulation. However, this error is low compared to other silicone formulations. The compressive response at dynamic strains must be similar to that of organic tissue, as dynamic strain rates (4.7 – 5.6 $m.s^{-1}$) are seen in rugby impacts.

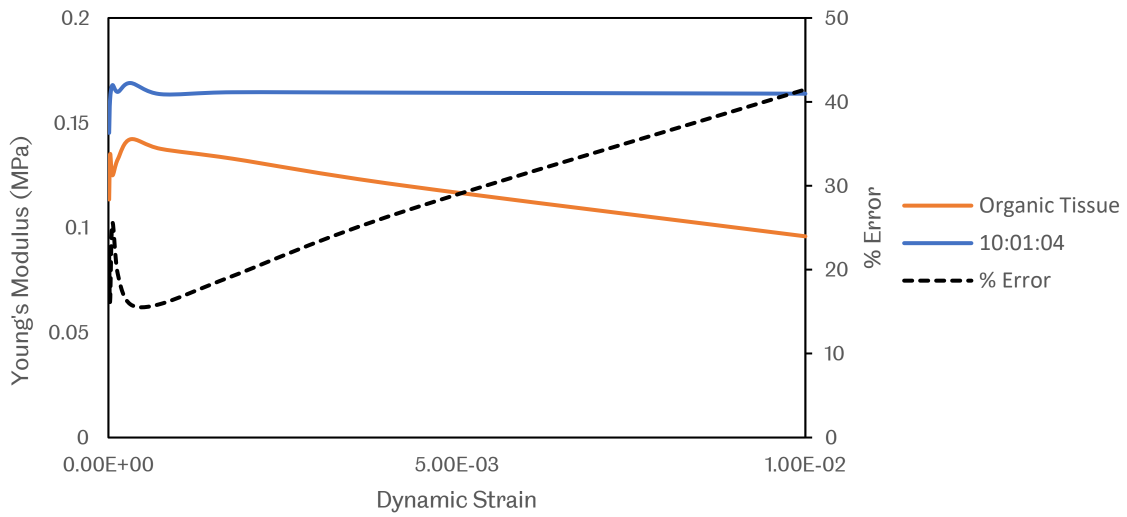


Figure 5.13 – Dynamic strain sweep young's modulus plot of relaxed porcine muscle tissue, 10:1:4 silicone formulation with % error overlayed.

5.3.6 Consistency of Response

The consistency of the silicone response was explored to examine the potential variability between different batches of silicone formulation 10:1:4 and also how the silicone's mechanical response may change over time. Five identically sized samples were fabricated, and their quasi-static compressive response was tested at a strain rate of 50mm/min using the same procedure described in §5.3.1. The results are detailed in Figure 5.14a with the compressive stress at different strains detailed in Table 5.5. To test for differences in the silicone over time, a sample was tested for its compressive response at 50 mm/min then tested again 30 days later, the results are detailed in Figure 5.14b, with the compressive stress at different strains detailed in Table 5.6.

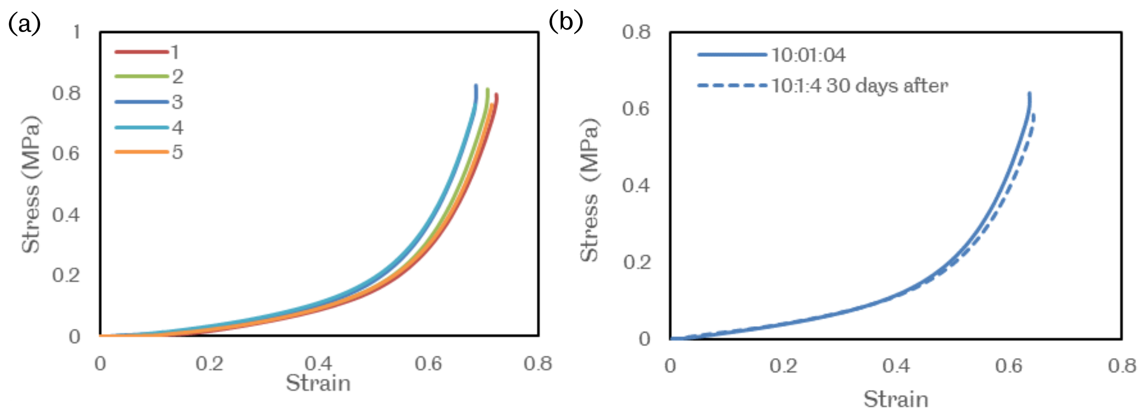


Figure 5.14 – Engineering stress-strain plots of (a) four identical 10:1:4 silicone samples (b) the same 10:1:4 silicone test sample tested 30 days apart.

The results show a small variance between each sample with SD becoming larger at each strain interval. The maximum differences from the experimental mean were 30.6 %, 12.8 % and 13.3 % at 0.2, 0.4 and 0.6 strain respectively. This shows a small variance in compressive response between sample to sample. This could be due to measuring inaccuracies as well as an inconsistent mix.

Table 5.5 – Compressive stress for four identical test samples as differing strains.

	Compressive Stress (KPa)		
Sample number	*0.2 Strain*	*0.4 Strain*	*0.6 Strain*
1	20	89	300
2	24	96	320
3	29	103	378
4	33	111	381
5	22	92	301
Mean	26	98	336
SD	6	9	40
Max. Difference from Mean	30.6 %	12.8 %	13.3 %

Stress-strain plots of the samples tested 30 days apart show a much smaller variance. The maximum differences from the experimental mean were 2.37 %, 0.235 % and 5.391 % at 0.2, 0.4 and 0.6 strain respectively. This suggesting the silicones' compressive properties do not change over time.

Table 5.6 – Compressive stress for the same test sample 30 days apart as differing strains.

	Compressive Stress (MPa)		
Sample	*0.2 Strain*	*0.4 Strain*	*0.6 Strain*
10:1:4	38	113	434
10:1:4 30 days after	040	114	388
Mean	39	114	411
Max. Difference from Mean	2.370 %	0.235 %	5.391 %

5.4 Simplified Shoulder Surrogate

5.4.1 Introduction

Section 5.4 presents the design and fabrication of a simplified shoulder surrogate, this has been informed by the anatomical and mechanical assessments of the human shoulder (§4), as well as the simulant material development previously outlined (§5.2). The design and manufacturing techniques used have been documented as well as justification for the simplification of geometries given. When developing a surrogate for injury biomechanics research, the range of structures modelled, the shape of the surrogate as well as the materials used need to be considered. These considerations are nominally affected by the surrogate's purpose. This purpose will then drive the appropriate levels of anatomical simplification of the surrogate. It is required that the surrogate can be used in a test house but also be as replicable of a human shoulder as possible so that viable conclusions can be made when testing padded clothing.

For the surrogate to be deemed appropriate for a test house it must give repeatable results. This means if the surrogate is repeatedly impacted with the same load, the output response should be similar. The surrogate must also be fabricated easily and in a manner that is replicable to the previous surrogate. These considerations will impact the surrogate's level of complexity. Many impact surrogates to date only use a single-layered approach to replicate the soft tissue made up in a body segment, more recent research has indicated using a multi-layered approach can provide a more biofidelic representation of the body segments response to impact [210]. Therefore, bone, muscle, subcutaneous adipose, and skin structures should be considered in the surrogate. The surrogate was developed in a way that it can:

- Produce a biofidelic response to impact representative of the human shoulder to ensure better assessment of padded rugby clothing, with a focus on Laceration, Cut and Abrasion injuries.
- Produce repeatable results so that they can be incorporated into test house use.
- Be fabricated easily and in a repeatable manner for use in a test house.
- Be durable to repeated impacts to extend its impact capacity.
- Be fabricated cheaply so that when destroyed, a new surrogate could be produced.

Therefore, the aims and objectives were to:

1. To develop a multi-layer shoulder surrogate using simplified geometries and soft tissue simulant materials.
2. Ensure fabrication of the surrogate was feasible and its response to impact was validated.

5.4.2 Methodology

a) Shoulder Surrogate Design

Results from the anatomical assessments of the shoulder as well as the intended purpose of the shoulder surrogate will affect its design, this is explained in §5.4.1. 3D scans of the shoulder show the closest common shape it represents is a half-cylinder (§4.2). Using this shape in the shoulder surrogate will allow it to be repeatable as well as easy to fabricate, a key need for test house use. Ultrasound scans on the shoulder showed 2 key layers above the clavicle, the trapezius muscle, and a skin and adipose layer (§4.3). A layered approach will be taken in the surrogate design to embody the key soft tissue structures of the shoulder. Layer thicknesses of soft tissues vary across the top of the shoulder from the trapezius (muscle thickness = 17 mm) along to the AC joint (muscle thickness = 0 mm). However, to make the surrogate appropriate for test house use, the layer thicknesses must be consistent throughout, therefore a 10 mm thick muscle layer was picked as a midpoint between the trapezius and the AC joint. Below these layers, a rigid element is needed as a basic representation of the rigid structures below the muscle (Clavicle and Scapula). Using these assessments and simplifications an appropriate shoulder surrogate was designed using Computer Aided Design (CAD) software Solidworks (Solidworks 2018, Dassault Systèmes, France) (Figure 5.15). The engineering drawings can be found in appendix B. The geometrical measurements and simplifications can be seen in Table 5.7.

Table 5.7 – Shoulder surrogate design measurement parameters.

Surrogate diameter (mm)	Surrogate length (mm)	Thickness (mm)		
		Bone	Muscle	Skin + Fat
105	15	37.5	10	5

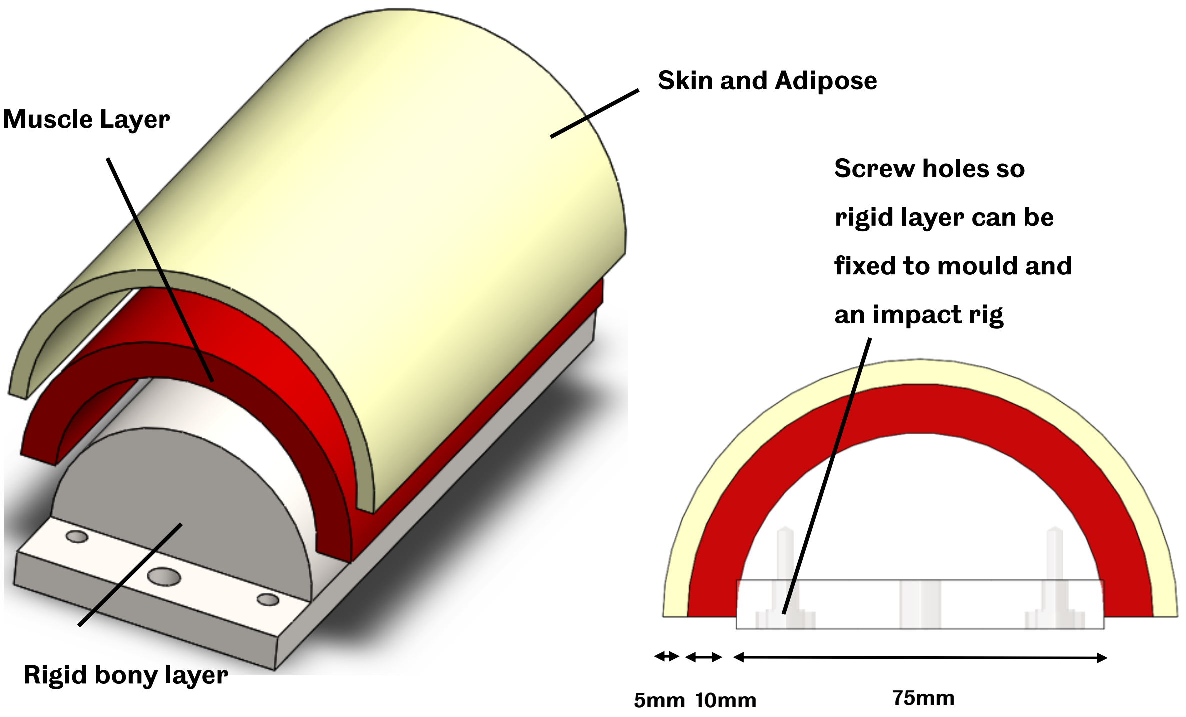


Figure 5.15 – Simplified shoulder surrogate CAD design.

The surrogate design was overlayed into the 3D scans of rugby players' shoulders (Figure 5.16) described in §4.2. This was to ensure its geometries matched the external geometries of a rugby player's shoulder closely. When observing these images, the use of a simplified cylindrical design can be warranted.

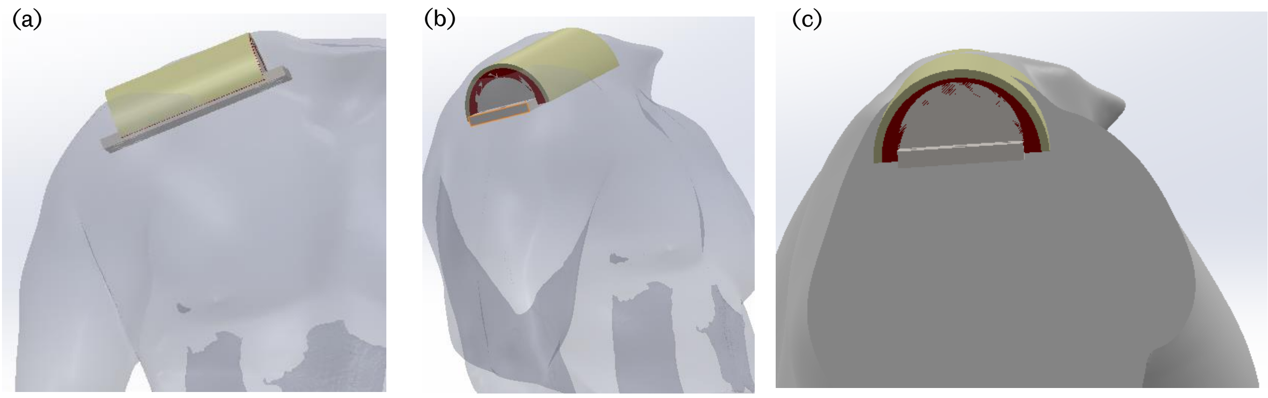


Figure 5.16 – Simplified shoulder surrogate design overlayed on the human body (a) Coronal plane, (b) Sagittal plane, (c) sagittal plane cut out of the shoulder.

The materials used in the surrogate must also be considered in the design process. They must display a similar response to load to the human tissue they embody as well as be repeatable, durable, and easy to fabricate. The bespoke silicone outlined in §5.3.5 was used as a relaxed muscle simulant. A commercially developed tissue plate (Syndaver®, Florida, Basic tissue plate) was selected as the skin (2 mm) and adipose tissue (3 mm) layer. The tissue plate is made from salt, water, and fiber, it has been fabricated and validated by Syndaver® to exhibit similar mechanical properties to their human equivalent including tensile modulus, abrasion resistance, penetration force, and coefficient of friction [211]

b) Shoulder Surrogate Fabrication Process

Development of the Mould

General Considerations

The three soft tissue components (skin, adipose, muscle) needed to be formed over a rigid layer representing the rigid structures under human soft tissues. To do this the following requirements for the mould were as follows:

- The mould will allow for silicones to be set into an accurate geometric representation of the shoulder surrogate design.
- Silicones are easily poured into the mould to ensure work time is kept low.
- Mould allows for a clear flow when pouring silicone to ensure mixing consistency and stop voids. This will create consistent material properties throughout the surrogate.

Mould Design

After many design iterations, it was decided that the rigid element of the shoulder surrogate would be used as an integral part of the mould, this ensured the silicone muscle layer set to an exact geometry. A CAD image of the mould is seen in Figure 5.17. The rigid element of the mould was fabricated out of steel in two parts, then screwed together, the reasons for this are listed:

- Steel has a similar compressive strength to the bone of 150 MPa [212].
- Steel has a density of 8000 kg/m^3 and impact toughness of 450 J/cm^2 (Charpy notched impact test) so will provide a suitable rigid element when placed in an impact rig set up.
- Steel is relatively easily machined with a smoothed finish to allow for more accurate moulding.

The mould base was fabricated from aluminium in three parts, then screwed together, the reasons for this are listed:

- Aluminium is both cheaper and lighter than steel.

- Aluminium is easily machined with a smoothed finish to allow for more accurate moulding.
- The mould could be taken apart so that when cured, the silicone layer could be easily removed.

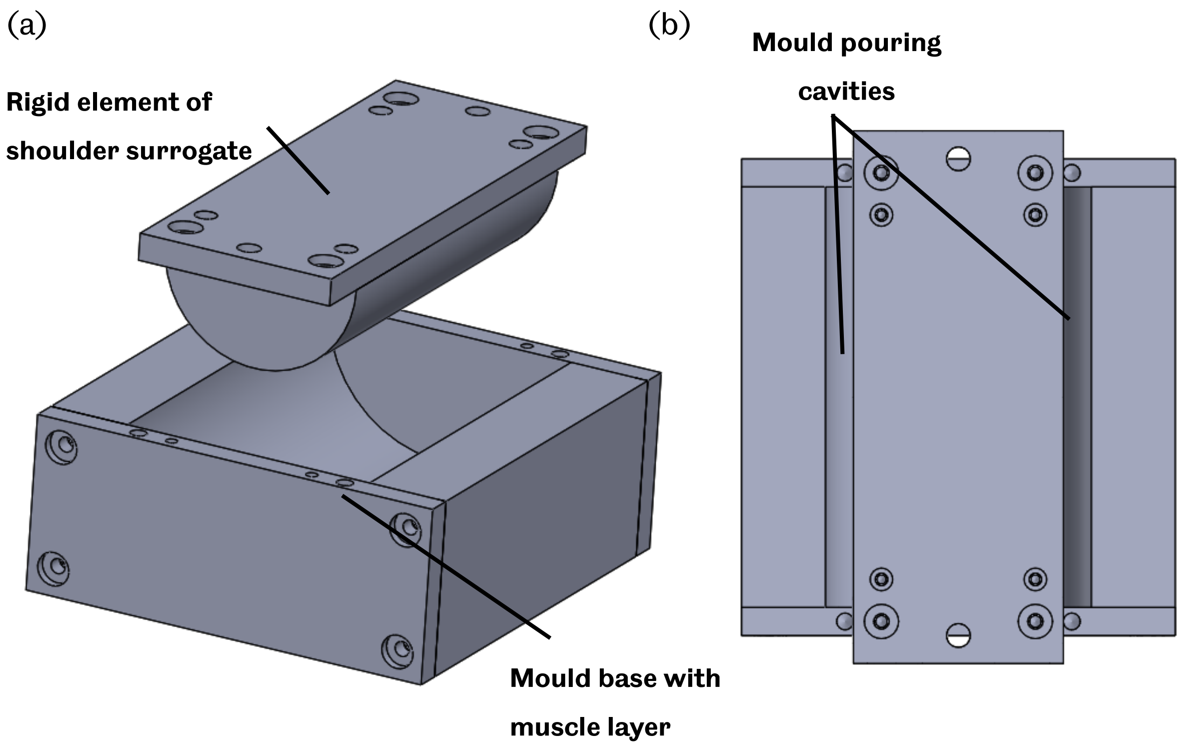


Figure 5.17 - Mould assembly with highlighted features (a) Exploded view, (b) Top view.

The final mould design allowed for inexpensive manufacture, accurate surrogate geometries, and ease of pouring silicones into the mould. Due to the rigid element of the mould also incorporating the bone part, the mould base could be made in different sizes to allow for different soft tissue layer thicknesses if needed.

Fabrication Procedures

a) *Muscle Layer*

The muscle layer used was formulated and mixed following the procedures outlined in §5.2.3. This was moulded independently to the skin and adipose layer. The procedures after formulating the silicone are listed:

- Mould is coated with release agent (Rocol, Mould Release Agent, UK).
- Silicone formulation is poured into moulds immediately after being degassed.
- Silicone specimen is left at room temperature for 24hrs to cure (Figure 5.16a)

- Each element of the mould base is screwed apart.
- Silicone specimen is removed from the mould and cleaned of any contaminants (Figure 5.18b)

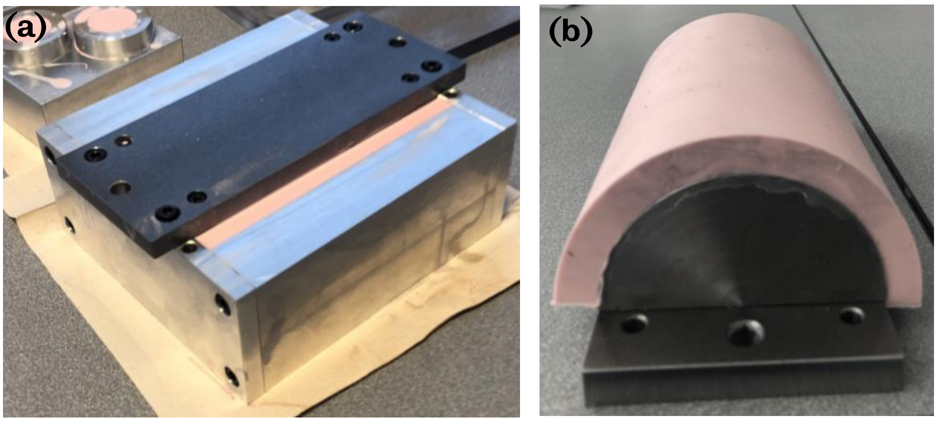


Figure 5.18 – Muscle layer moulding process (a) Silicone muscle layer moulded around the rigid element of surrogate, (b) Cured muscle layer removed from mould.

b) Skin and Adipose Layer

The skin and adipose layer were supplied commercially (Syndaver®, Florida, Basic tissue plate) due to its similarities in compressive response and puncture resistance to organic tissue. The tissue plate constituted of a 2mm skin layer and 3mm adipose layer. The layer was cut to the correct size (15 x 16.5 cm) and then stuck with an adhesive spray (Ambersil, England, HS 300) to the muscle layer to create a 4-layer surrogate (Figure 5.19).

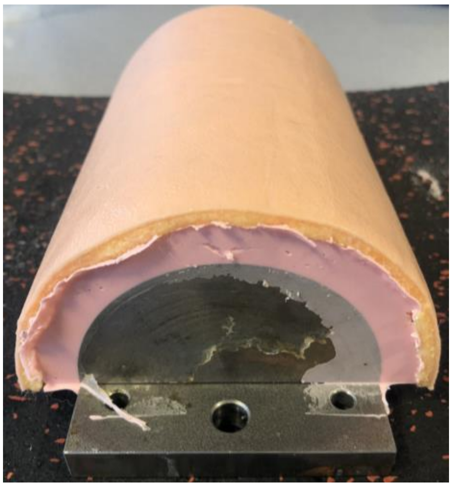


Figure 5.19 – Skin and adipose layer moulded to the rigid bone element of surrogate.

c) Final Surrogate Fabrication with Syndaver® Skin and Adipose Layer

After all processes are completed, the final fabrication was complete. A simplified semi-cylindrical geometry representative of the superior part of the shoulder was produced. The shoulder surrogate includes a steel core as a basic representation of the rigid structures below the muscle (Clavicle and Scapula), a silicone muscle layer, and an outer skin + adipose layer (Figure 5.20). Adhesion issues between the muscle layer and skin and adipose layer were seen.

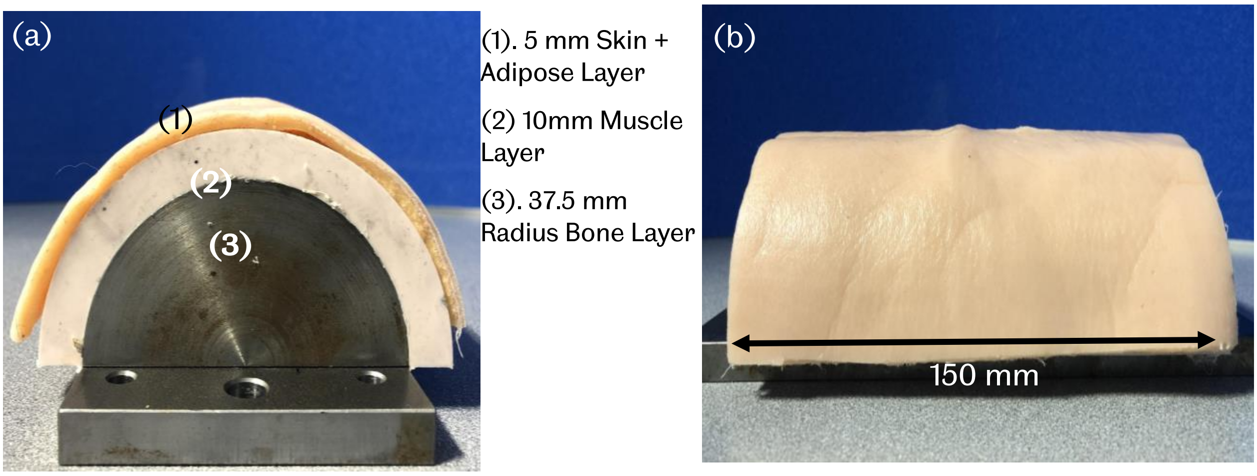


Figure 5.20 – Simplified Shoulder Surrogate (Syndaver® skin and adipose layer) (a) cross-sectional view, (b) side view.

d) Use of Synthetic Chamois Skin

The final simplified surrogate outlined above used a state-of-the-art, synthetic skin and adipose layer produced by a commercial supplier (Syndaver®, Florida, Basic tissue plate). Although this provided a pre-validated outer layer to the surrogate, the use of this caused some key issues:

- When damaged, the surrogate could not be impacted in this location again.
- Bonding the skin and fat layer to the silicone muscle layer was difficult due to the type of materials used in each layer.

It was set out in the design criteria that the surrogate needed to be inexpensive to re-fabricate when damaged. Due to the nature of the research project, the shoulder surrogate would be impacted in conditions that could cause a tear to the skin and fat layer. An alternative skin and adipose option needed to be explored so that cost could be kept to a minimum, and the skin and adipose layer could be bonded more efficiently to the muscle layer. The following requirements were set out to develop a shoulder surrogate with a different skin and adipose layer:

- The skin layer must be inexpensive to keep fabrication costs to a minimum.
- The skin layer must be easily bonded to the muscle layer.
- The skin layer must have similar puncture resistance to organic tissue.

Chamois leather has been used in previous sports and ballistics research to simulate the skin because of their similarities in puncture resistance [125, 126, 157, 158]. It is also relatively inexpensive and easy to mould around a cylindrical shape. Due to this, a chamois skin was used. The muscle and adipose tissue layers were combined to create a 'soft tissue layer' due to their similar compressive properties (Table 4.6) and the minimal amount of adipose found in the human shoulder.

a) Updated Mould

A very similar mould to the simplified surrogate with Syndaver® skin and adipose was used. However, this mould was slightly larger (plus 10 mm width), as the chamois skin would sit in the mould when curing took place. Combining the muscle and adipose layer also needed to be considered. The mould allowed for a 15mm cavity between the rigid bone element and mould surface (Figure 5.21).

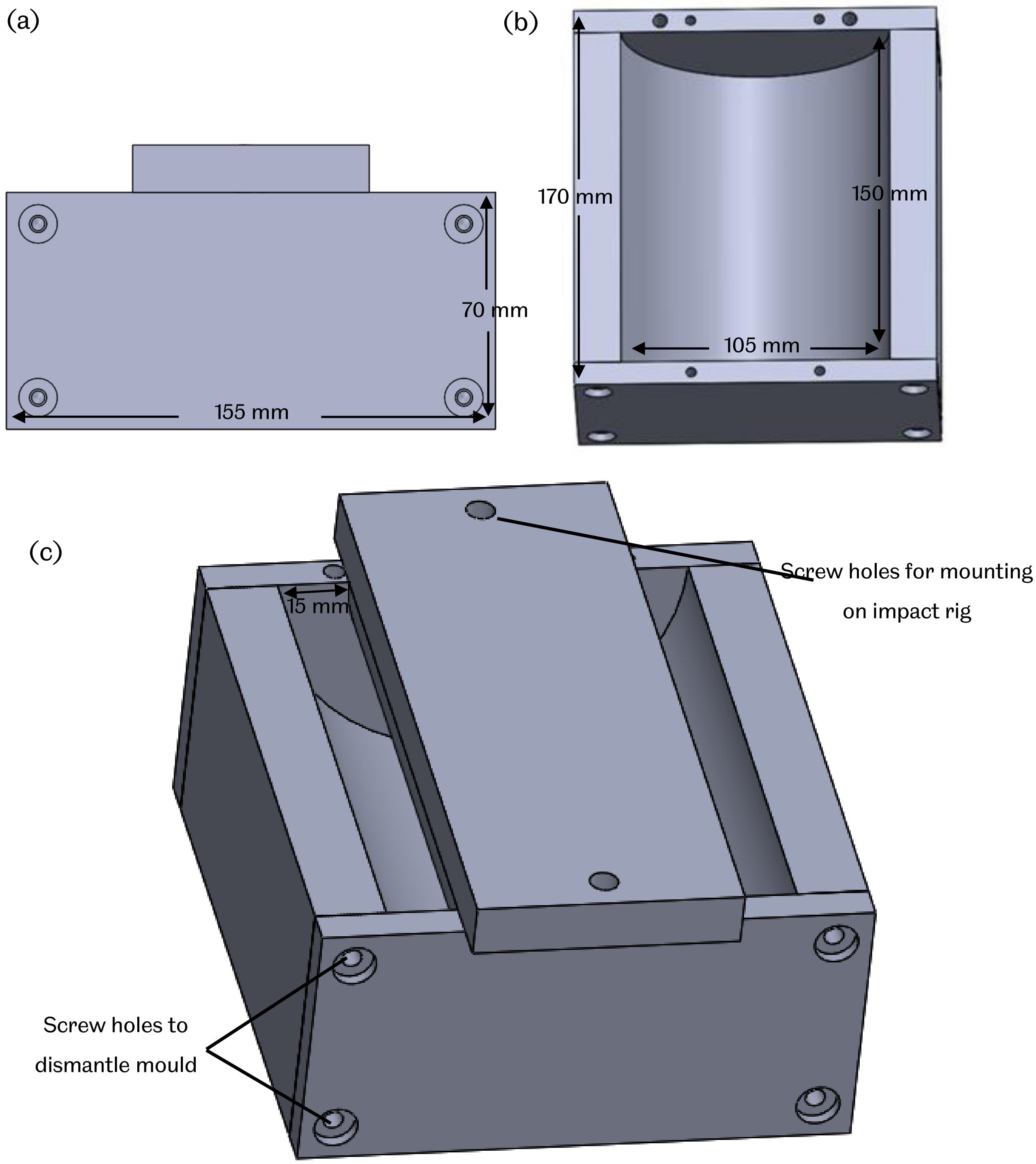


Figure 5.21 – Mould assembly with highlighted features (a) side view, (b) superior view without rigid bone part, (c) angled superior view with the rigid bone part.

b) Fabrication Procedures

A similar process was used to fabricate the surrogate with chamois skin. However, only one moulding step was needed. The procedures are as follows:

- 2mm thick chamois leather skin (KCIC200, Kent Car Care, Manchester, UK) cut to 150 mm length x 160 mm width and placed in the bottom of the mould (Figure 5.22a)
- The mould and rigid element are coated with a release agent (Rocol, Mould Release Agent, UK).
- The Silicone (soft tissue) layer is formulated and mixed following the procedures outlined in §5.2.3.
- Silicone (soft tissue) formulation is poured into the mould immediately after being degassed (Figure 5.22b).
- Silicone specimen is left at room temperature for 24hrs to cure (Figure 5.22b)
- Each element of the mould base is screwed apart.
- Silicone specimen is removed from the mould and cleaned of any contaminants.

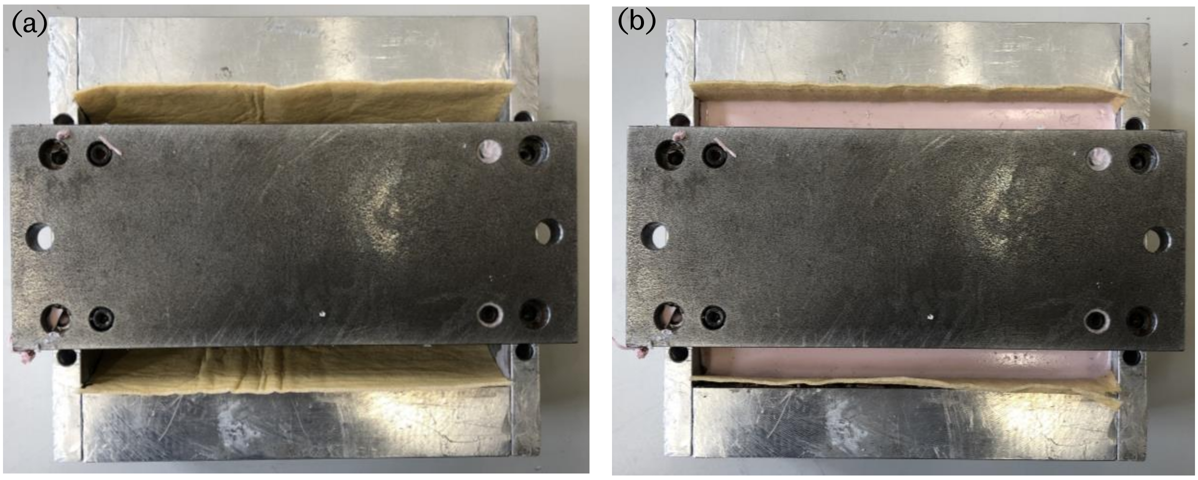
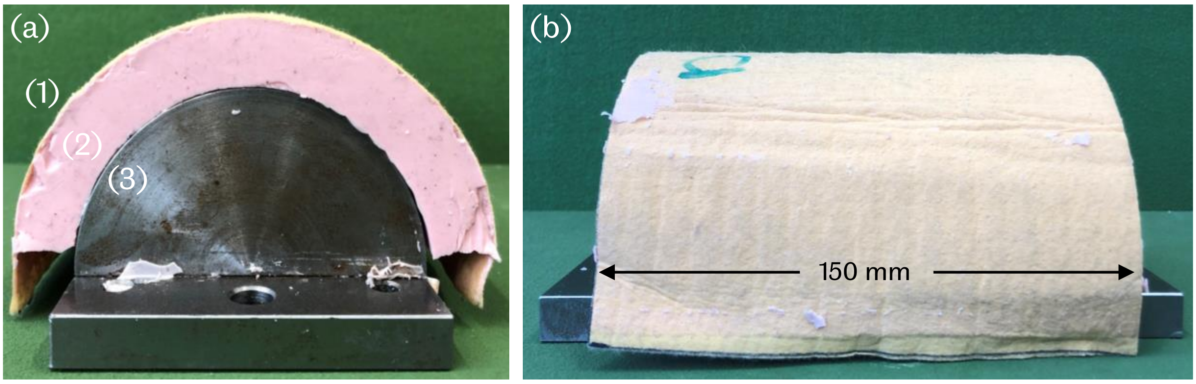


Figure 5.22 – Moulding process – (a) skin layer, (b) skin and soft tissue layer.

c) Final Surrogate Fabrication with Chamois Skin Layer

After all moulding processes are completed, the final fabrication was complete. The simplified shoulder surrogate had a 2 mm thick chamois skin later and a 13 mm thick soft tissue layer (muscle and adipose) (Figure 5.23). These fabrication procedures meant the final simplified surrogate was inexpensive and repeatable to produce.



* (1) – 2 mm skin layer, (2) - 13 mm soft tissue layer, (3) - 37.5 mm Radius Bone Layer

Figure 5.23 – Simplified Shoulder Surrogate (chamois skin layer) (a) cross-sectional view, (b) side view.

5.4.3 Cost

The cost of manufacture for the simplified shoulder surrogate is required to be affordable due to its intended use. The simulation of cuts and Lacerations will destroy the surrogate meaning a new one would have to be manufactured. A full cost breakdown for one simplified shoulder surrogate is outlined in Table 5.8.

Table 5.8 – Cost breakdown for one simplified shoulder surrogate.

Item		**Cost Per Unit**	**Quantity**	Cost (£)
Soft Tissue Layer Constituents	Silastic™ 3481	£15.36/kg	**0.4kg**	6.14
	Deadener	£33.60/kg	**0.2kg**	6.72
Skin Layer Constituents	Syndaver® basic Tissue Plate	£2200/m^3	**0.024m^3**	52.50
	Synthetic Chamois Leather	£16.62/m^3	**0.024m^3**	0.40
Steel Rigid Bone Element				75.00
Aluminium Mould				53.00
Laboratory Consumables				10.00
Service Costs (Machining, Degassing)				20.00
			Total Cost (Syndaver® skin)	£223.36
			Total Cost (Chamois skin)	£179.86

The total cost for the simplified shoulder surrogate was £223.36 (Syndaver skin) and £179.86 (Chamois skin) respectively. This is the initial cost that occurred to develop an entirely new surrogate. Both the bone element and the mould can be reused for following fabrications. Considering this, and the use of a chamois skin, the cost of the following surrogates would be greatly reduced to £51.86 per surrogate. This would therefore enable a cost-effective approach for use an impact surrogate. The number of impacts before the surrogate is destroyed would have to be considered. Costs could rise if the surrogate was destroyed after every impact.

5.4.4 Compressive Properties of Simplified Shoulder Surrogate

The surrogate's compressive properties at a quasi-static loading rate were established so that:

- This could be compared with data from other surrogates.
- The repeatability of the surrogate could be measured.

A Shimadzu mechanical test machine (Shimadzu, EZ-LX, Kyoto, Japan) was used to perform indentation tests on the simplified shoulder surrogate with chamois skin. A flat indenter with a diameter of 16 mm was used at a loading speed of 5mm/ min. Three indentation tests were completed at different locations down the top of the surrogate (Figure 5.24a). The test was stopped at 10 mm displacement from the surface of the surrogate. Similar force traces were seen in each test (Figure 5.24b).

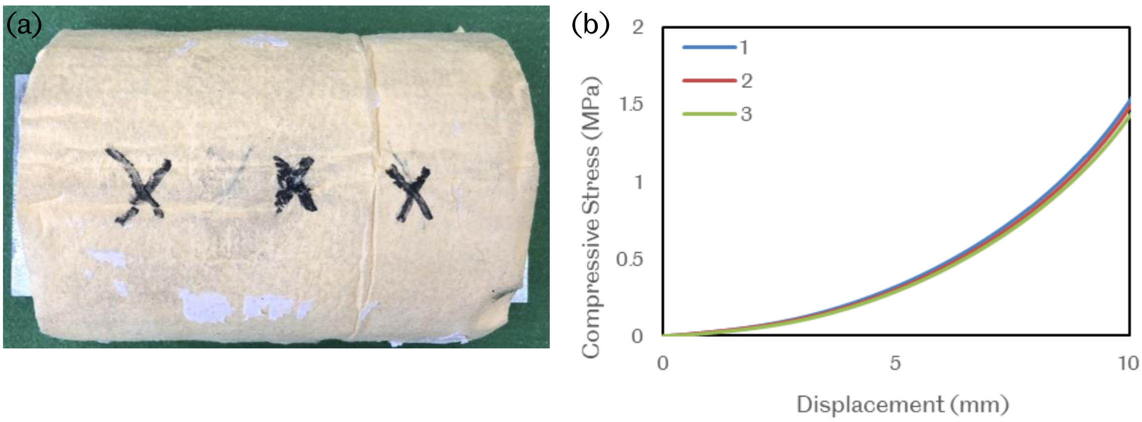


Figure 5.24 – (a) Surrogate with indentation locations marked, (b) Compressive Stress-Displacement trace for three surrogate locations (5 mm/min).

There was a good consistency of response throughout the surrogate with a maximum percentage difference from the mean being 9.66 % at 2 mm displacement (Table 5.8). This suggests the surrogate will give a repeatable response to impact, this is crucial for the consistent testing of sports padding.

Table 5.9 – Compressive stress at differing locations on the simplified shoulder surrogate at differing displacements (mm)

	Compressive Stress (MPa)		
Sample number	*2 mm*	*5 mm*	*10 mm*
1	0.063	0.320	1.521
2	0.060	0.303	1.479
3	0.053	0.290	1.421
Mean	0.059	0.304	1.474
SD	0.005	0.015	0.050
Max. Difference from Mean	9.66%	5.15%	3.57%

5.4.5 Discussion

Quality of Simplified Shoulder Surrogate

The simplified shoulder surrogate aimed to design and fabricate a human shoulder surrogate, for use in impact testing of shoulder padding. Although many anatomical simplifications were made, it is intended that the surrogate's dynamic response to impact will more closely match what is seen in the human equivalent when compared to what has been presented to date. Both a four and three-layer surrogate, made of skin, adipose, muscle, and rigid bone structures taking into account their representative layer thicknesses on a real human. This is much more representative than a rigid cylinder which is used as a shoulder impact surrogate for the testing of many padded clothing products in the sports industry. The anatomical geometries of the surrogate were determined through both 3D and Ultrasound scanning of an appropriate data set of rugby players, which links to the surrogate's application to be used in the impact testing of rugby shoulder padding. No other research has used an anatomical dataset that matches its application. Payne [210] used a data set from Visible Human Project (VHP) [213] and outlined that this was not representative of an athletic population. Establishing how a rugby player's geometry differs from the VHP dataset would be beneficial for the future design of sports impact surrogates. The surrogate's primary use is to be used in impact testing of rugby shoulder padding. The impact surrogate must be replicable, durable, and repeatable [11]. The simplification of certain geometries in the surrogate has made this achievable. The surrogate could therefore be used in comparison with simpler surrogates that could be used in a test house. The addition of a skin layer made it possible to replicate Laceration, Cut and Abrasion injuries.

Coupled with the uniformity in the surrogate's structure, repeatable and accurate testing could be achieved.

It can be assumed the muscle layer would affect the surrogate's response to impact due to its thickness and relative stiffness [210]. The compressive response to load was replicated to Porcine muscle tissue, novel research was obtained to characterise porcine muscles compressive response. It has been identified that many factors including specimen age, site, and preparation [109] can affect these characteristics. Obtaining novel data means this can be controlled as well as adding to current organic tissue data. Low load testing on human subjects should be completed to further validate the shoulder surrogate.

The fabrication approach also enabled a cost-effective surrogate that was suitable for repeat testing of shoulder padding as well as the use as a frangible surrogate for the simulation of Laceration injuries. The total cost of the surrogate with chamois skin was £179.86, and identical surrogates could be made for repeat testing at a cheaper cost of £51.86. Payne [168] developed a human thigh surrogate costing £2441, far more expensive than the surrogate developed here. It must however be noted that fewer anatomical simplifications were made, and the human thigh is larger than the shoulder meaning more simulant materials were needed.

Simplified Shoulder Surrogate Improvements

Many geometrical simplifications have been made to the shoulder surrogate for both repeatability and ease of fabrication. How this may affect the surrogate's response to load is somewhat unknown. The shoulder embodies a half-cylinder with 4 tissue layers (3 on chamois surrogate) through this, as represented in the current surrogate. However, data obtained from both the research presented and from other data sets, namely the VHP [213] shows that the geometries of these layers differ throughout the shoulder. Future surrogates should look to use these datasets to further match the 3D tissue morphologies of the shoulder. Data like the VHP must be used with caution as it is not representative of an athletic population. Adipose tissue thickness could be largely reduced with muscle tissue thickness being largely greater. Coupled with this the range of body compositions seen in rugby players from position to position makes replicating their tissue structures difficult. Further imaging technologies like magnetic resonance imaging (MRI), magnetic resonance arthrography (MRA), and CT arthrography (CTA) [214] could be used to create a more defined athletic dataset to model a future shoulder surrogate. Contrary to this, both the female and youth populations should be considered as they are regular users of shoulder padding also [215]. Technologies like 3D printing as well as computer-aided machining of the mould for the surrogate could be used to fabricate a more anatomically representative surrogate. However, both the surrogate's repeatability and its ease of fabrication would be lost. Although a more biofidelic surrogate could be constructed, its final application must be considered.

Although the surrogate uses materials that replicate the compressive response of their human tissue counterpart, the surrogate has not been fully validated with data from humans *in vivo.* It would not be ethically viable to obtained compressive stress-strain data at high loads. However, compressive stress-strain data could be gathered at loads that do not cause pain to a human.

The effect of muscle tension on the compressive load response of organic tissue was not considered for this study. Many biomedical studies have investigated the effect of muscle contraction on its mechanical characteristics [216-218]. However, current literature provides no conclusive data on how levels of contraction can change the response of the tissue. Tackle biomechanics research shows the trapezius muscle is contracted when making a rugby tackle [61]. When considering the application of this shoulder surrogate, further research is required in the development of contracted muscle simulants.

5.5 Anatomical Surrogate

5.5.1 Introduction

Chapter 5.5 presents the design and fabrication of an anatomical shoulder surrogate that aims to represent more closely the geometries in a rugby player's shoulder compared with the simplified surrogate presented in §5.4. The development is informed by the anatomical and mechanical assessments of the human shoulder (§4), as well as the simulant material development previously outlined (§5.2). The design and manufacturing techniques used have been documented as well as justification for the simplification of geometries given.

The purpose of the surrogate was to be used in impact testing so the impact response of both the shoulder and padded clothing in differing impact locations could be explored. Unlike the simplified surrogate, it was therefore required the external and internal geometries closely represented its human counterpart. There was no intention to use this surrogate in a test house, so no considerations towards its repeatability needed to be made. Considerations were made to the surrogate's complexity taking both technology and time in mind. However, a two-layer approach was taken in that a rigid bone layer and soft tissue layer in which both had matching external geometries to their human counterparts were developed. A large amount of the soft tissue in the shoulder is muscle tissue, therefore this was used to represent the soft tissue layer. A skin layer was not required as the surrogate was not being used to explore Cut, Laceration, and Abrasion injuries, only the loading response of impact. Coupled with this, technology and time made the development of a skin layer less possible. With these considerations in mind the surrogate was developed in a way that it can:

- Represent the external geometries of its human counterpart.
- Produce a biofidelic response to impact representative of the human shoulder to ensure better assessment of padded rugby clothing.

Therefore, the aims and objectives were to:

1. To develop a two-layer shoulder surrogate using representative anatomical geometries and soft tissue simulant materials.
2. Ensure fabrication of the surrogate was feasible and its response to impact was validated.

5.5.2 Methodology

Surrogate Skeletal Component

The skeletal components of the shoulder are key in influencing its mechanical response to impact. The shoulder is made up of three bones, the Scapula, Clavicle, and the Humerus. Human scans of these bones were exported from SketchFab into Solidworks and key geometries were measured. It was chosen that participant 1's geometries (§4.2) would be replicated as this was the median in the data set. Equations 2.1 – 2.5 displayed in §2.4.2 were used as a guide for the current bone dimensions (Table 5.10) and scaled to size using the scale tool in Solidworks.

Table 5.10 – Dimensions of appropriate bones used in the surrogate skeletal component.

Component	Dimensions (mm)
Humerus	
Length	364
Diameter of Humerus head	48
Scapula	
Transverse length	140
Longitudinal length	213
Clavicle	
Length	180
Acromial diameter	23
Sternal diameter	22

After the bones were imported and scaled in Solidworks, their origins were set to ensure they represented the human shoulder anatomy. A rigid base plate (261.0 x 195.8 x 20.0 mm) was developed so that that the bones were fixed. It was decided that this would be at the end of the Humeral head as this would mean the whole of the human shoulder would be included and any loading effects from deeper tissues could be considered. The rigid base plate also aided the moulding process described below. The clavicle would usually attach to the Sternum. However, the Sternum is not part of the human shoulder. Therefore, a 5 mm radius support was added to the surrogate to provide support to the clavicle in a way the Sternum would in the human body. The final CAD model is shown in Figure 5.25.

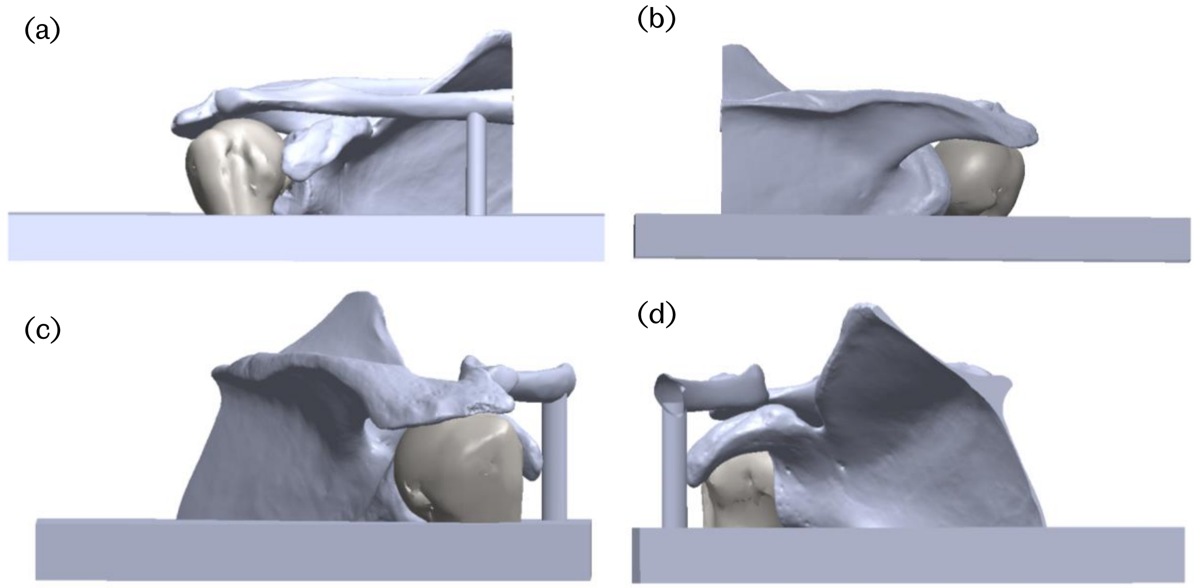


Figure 5.25 – CAD model of surrogate skeletal component (a) Anterior view, (b) Posterior view, (c) Lateral view, (d) Medial view. *(AC joint has been fused for to aid 3D printing).*

After a CAD model of the surrogate skeletal component was complete, fabrication needed to take place using 3D printing technologies. Considerations were made regarding the 3D printing technique used and appropriate materials representative of the mechanical response of cortical bone, while also taking into account cost and availability. Due to the cost and complexity, CNC milling the surrogate out of steel was not possible. Therefore, the surrogate skeletal component was laser sintered out of Glass Filled Polyamide (PA-GF). This material was readily available, and its mechanical properties were the closest to that of cortical bone when compared with any other technologies or materials (Table 5.11).

Table 5.11 – Mechanical properties of human cortical bone compared with PA-GF.

Mechanical Property	Cortical Bone	PA-GF
Density (g/cm^3)	1.47 – 2.12	1.22
Compressive Strength (MPa)	130 - 200	170 - 188
Compressive Modulus (MPa)	11.5 – 17	6.6 – 7.2
Tensile Strength (MPa)	50 - 151	51
Shore D Hardness	90	80

The final surrogate skeletal component is pictured in Figure 5.26
below.

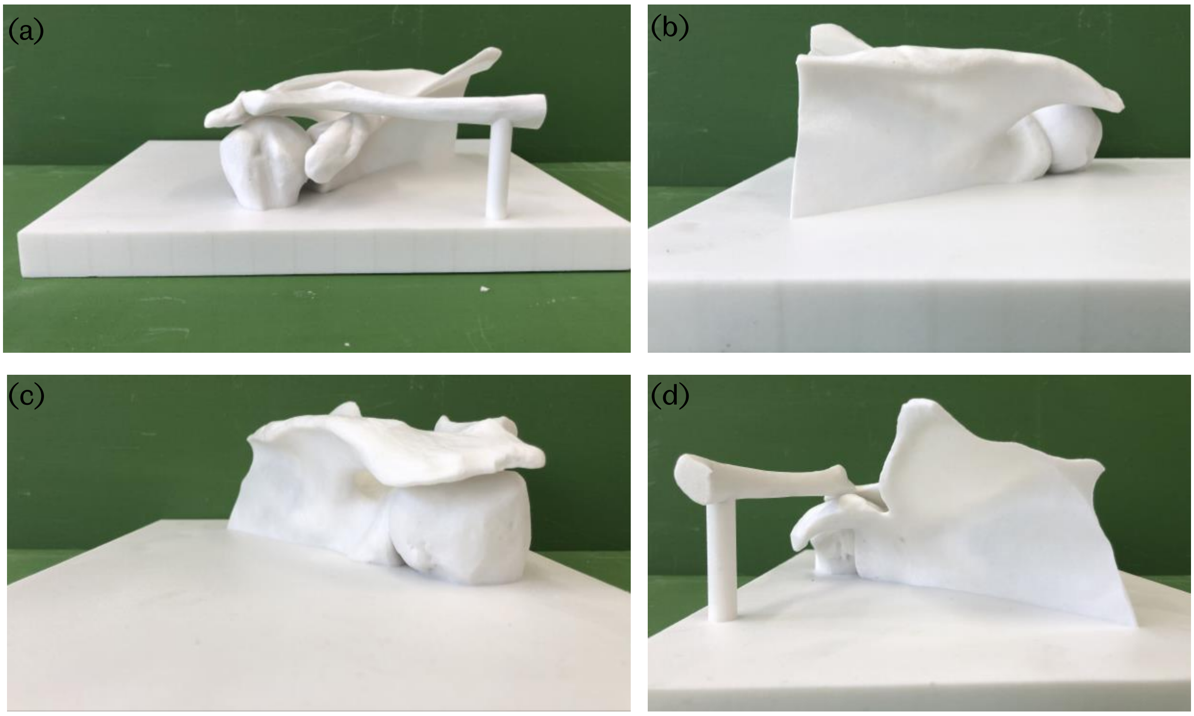


Figure 5.26 – Final surrogate skeletal component (a) Anterior view, (b) Posterior view, (c) Lateral view, (d) Medial view. *(AC joint has been fused for to aid 3D printing).*

External Mould Development

As stated in §5.5.1 the shoulder surrogate would be two-layered with a rigid bone layer and a soft tissue layer. When considering moulding, an external mould needed to be developed. However, some anatomical simplifications were made. The surrogate embodied a two-layer (skeletal and soft tissue layer) approach, skin, adipose tissue, and muscle layers were not considered. The soft tissue layer would represent relaxed muscle as this made up most of the tissue in the shoulder. Other tissues like cartilage and nerves were also not considered. Participants 1's (§4.2) external geometries from 3D body scanning were used as these were the geometries used for the surrogate skeletal component. The muscle layer needed to be moulded around the skeletal component, it was proposed that the same silicone developed in §5.2 would represent the muscle layer (10:1:4). Key requirements for the mould were established:

- The mould will allow for silicones to be set into an accurate geometric representation of the shoulder surrogate design, while allowing correct alignment of the skeletal component.
- Silicones are easily poured into the mould to ensure work time is kept low.
- Mould allows for a clear flow when pouring silicone to ensure mixing consistency and stop voids. This will create consistent material properties throughout the surrogate.

Using 3D CAD software Solidworks and participant 1's 3D scan of their external geometries a mould was developed by using the surface from mesh tool. This is pictured in Figure 5.27a. When developing the mould, a CAD model of the surrogate skeletal component was imported into the design so that the mould could be correctly aligned and ensure accurate geometries. Final CAD drawings of the mould are pictured in Figures 5.27c & 5.27d.

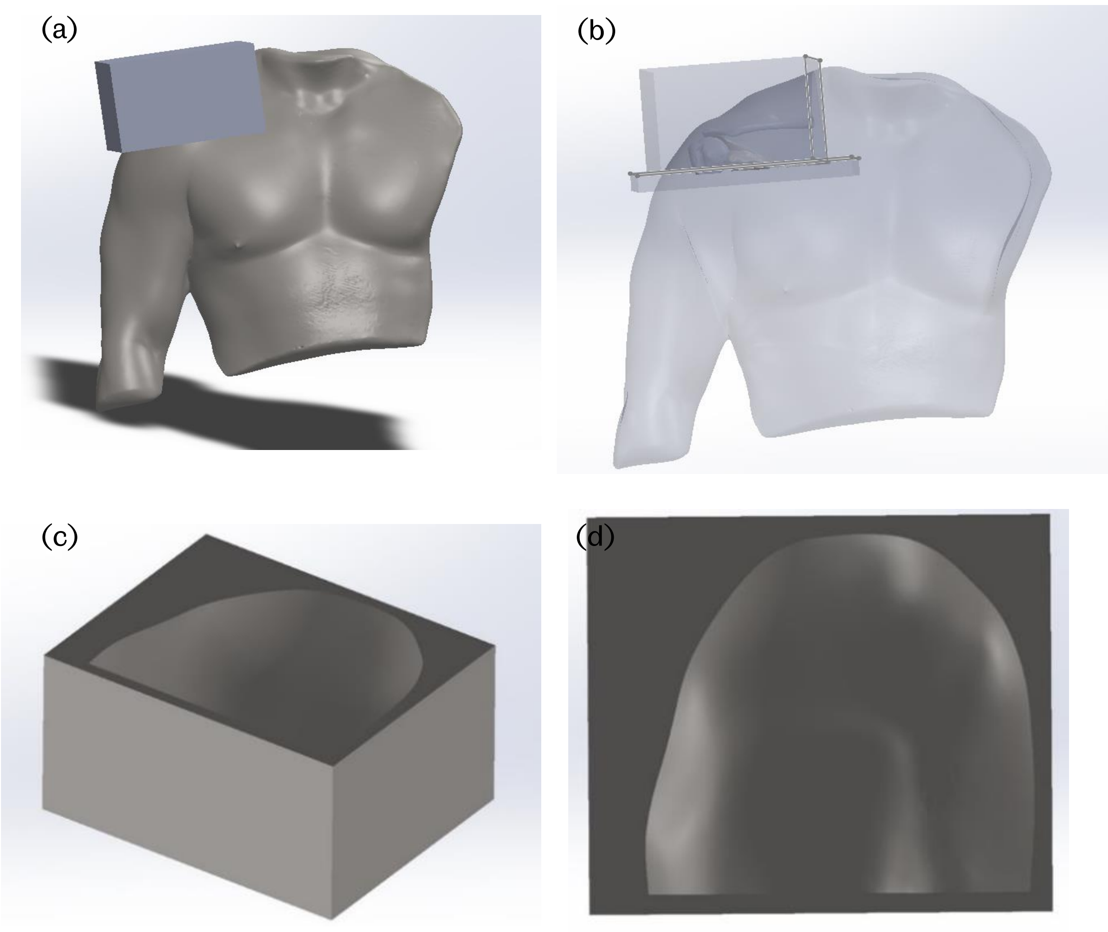


Figure 5.27 – Shoulder surrogate mould – (a) Overlayed on Participant 1, (b) Overlayed on Participant 1 (skeletal component showing), (c) Side view, (d) Overhead view.

After CAD drawings had been completed and mould designs were satisfied, the mould was 3D printed from Polylactic Acid (PLA) due to its stiffness and low cost. The printed mould is pictured in Figure 5.28 below.

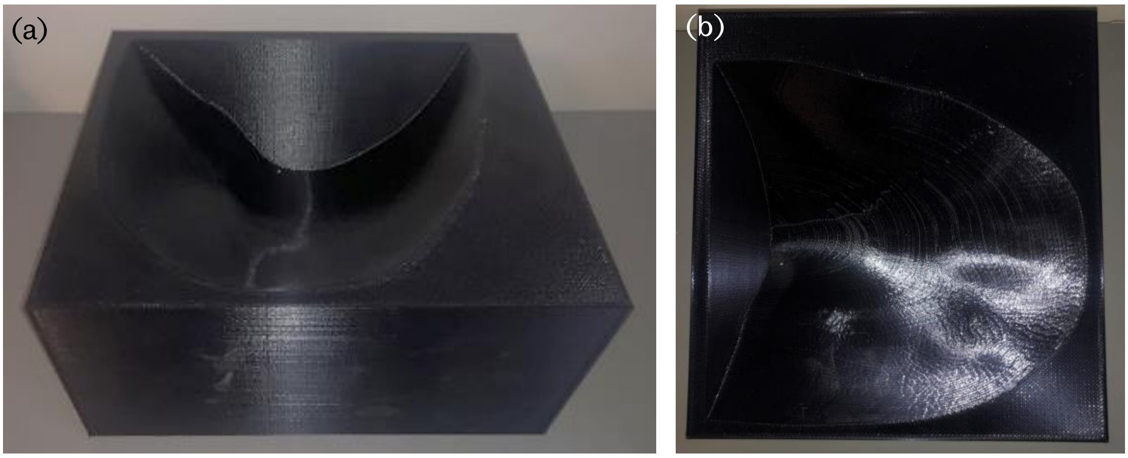


Figure 5.28 – Anatomical shoulder surrogate mould (a) Lateral overhead view, (b) Overhead view.

A final CAD design of the skeletal and muscle layer components was rendered to ensure their anatomies and geometries would be fully representative of a human shoulder. An annotated Figure is pictured below (Figure 5.29). The design featured a Clavicle support so the Clavicle was not 'floating' in the silicone layer. The Clavicle would usually attach to the Sternum, the design did not feature the Sternum, this was the best solution in keeping the Clavicle mechanically supported.

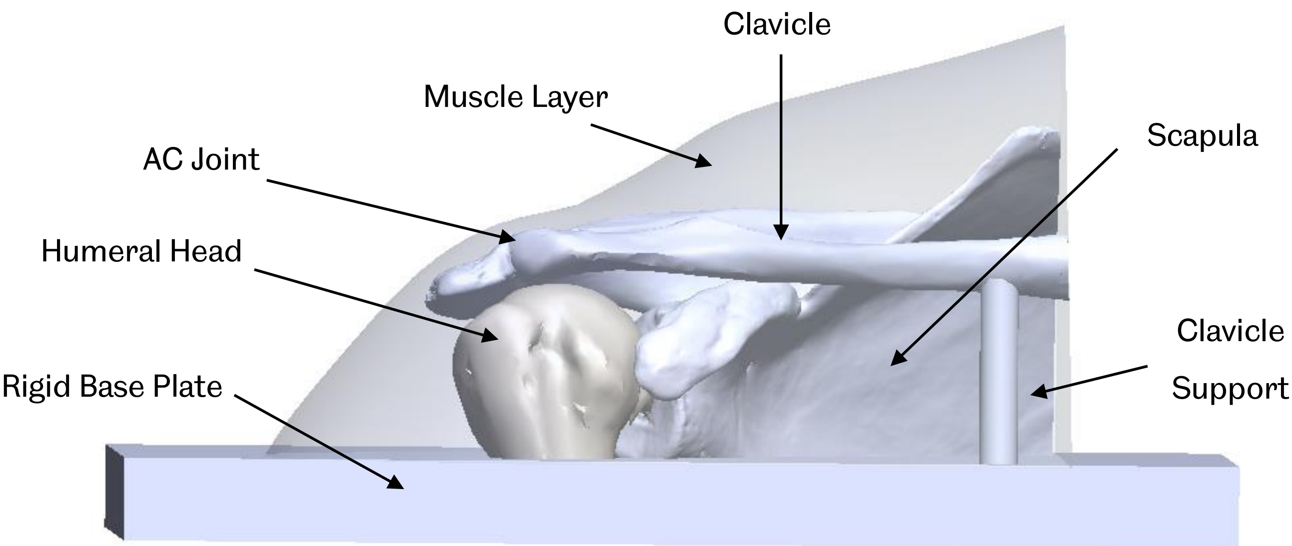


Figure 5.29 – CAD design of anatomical shoulder surrogate.

Surrogate Fabrication

The fabrication procedures for the shoulder surrogate were relatively simple and straight forward. Correct alignment of the skeletal component on the mould was crucial, this was done by aligning it with pre-cut reference points and then weighing the skeletal component down with a 10 kg mass (Figure 5.30a). The muscle layer was formulated and mixed following the procedures outlined in §5.2.3. The procedures after formulating the silicone are listed:

- The mould is coated with a release agent (Rocol, Mould Release Agent, UK).
- Silicone formulation is poured into the mould immediately after being degassed.
- Silicone specimen is left at room temperature for 24hrs to cure (Figure 5.30b).
- The shoulder surrogate is removed from the mould and cleaned of any contaminants.

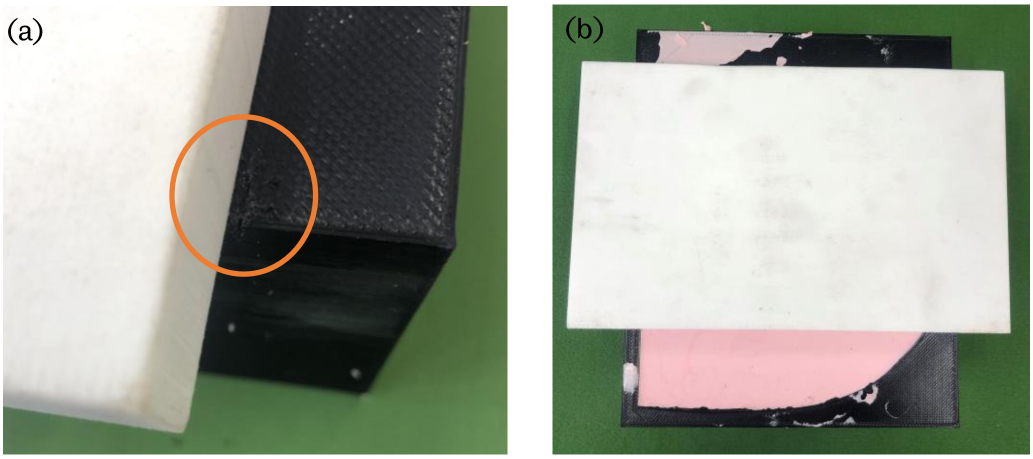


Figure 5.30 – Shoulder Surrogate moulding process (a) Pre-cut alignment guides (circled), (b) Silicone specimen in mould.

After all moulding processes are completed, the final fabrication was complete. A two-layer anatomical shoulder surrogate was produced with the silicone soft tissue layer set to the bone layer (Figure 5.31). The surrogate provided an accurate representation of external geometries, and the silicone was consistent throughout. It should be noted that the surrogates base did not extend across the whole soft tissue layer to aid fabrication and moulding as seen in Figure 5.30b.

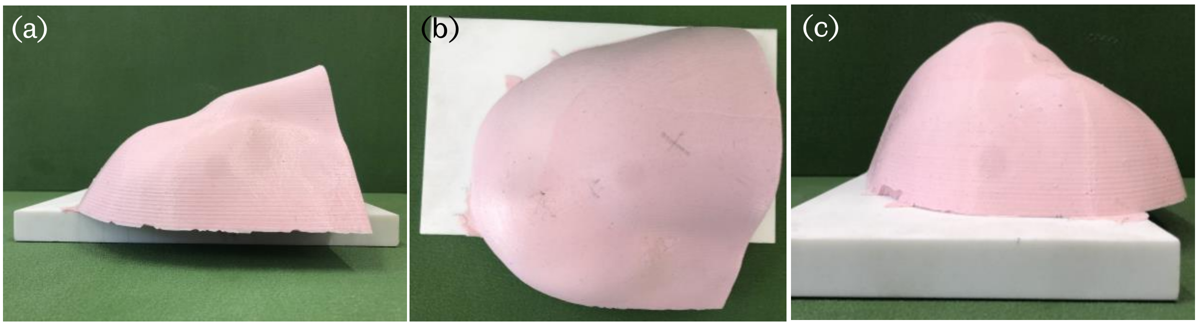


Figure 5.31 – Anatomical Shoulder Surrogate (a) Anterior view, (b) Superior view, (c) Lateral view.

5.5.3 Cost

The cost of manufacture for the anatomical shoulder surrogate was important to document. Although no requirements were made to keep the surrogate affordable, a comparison could be made to the simplified shoulder surrogate and any other human impact surrogates produced in the past or the future. A full cost breakdown can be seen in Table 5.12.

Table 5.12 – Cost breakdown for one anatomical shoulder surrogate.

Item		**Cost Per Unit**	**Quantity**	Cost (£)
Soft Tissue Layer Constituents	Silastic™ 3481	£15.36/kg	**0.6kg**	9.22
	Deadener	£33.60/kg	**0.3kg**	10.08
Skeletal Component				535.15
PLA Mould				200.00
Laboratory Consumables				20.00
Service Costs (Machining, Degassing)				20.00
			Total Cost	£794.45

The total cost of the anatomical shoulder surrogate was £794.45. This is the initial cost that occurred to develop an entirely new surrogate. Most of the cost occurred with the Skeletal component and PLA mould. The skeletal component can be reused unless fracture occurs during impact testing, and the mould can be reused for any other iterations required. This may occur if a new soft tissue layer with differing mechanical properties is developed. The cost of the anatomical surrogate was £614.59 more than the simplified surrogate, however, more biofidelic features have been achieved.

5.5.4 Compressive Properties of Anatomical Surrogate

The anatomical shoulder surrogate's compressive properties at quasi-static loading rates were established so that:

- Differences in load response between different locations on the surrogate could be measured.
- This could be compared with data from other surrogates.
- The repeatability of the surrogate could be measured.

A Shimadzu mechanical test machine (Shimadzu, EZ-LX, Kyoto, Japan) was used to perform indentation tests on the anatomical shoulder surrogate. A flat indenter with a diameter of 16 mm was used at a test speed of 5 mm/min. Tests were completed on three locations on the shoulder: the AC joint, the Trapezius insertion into the shoulder, and the middle belly of the Trapezius as pictured in Figure 5.32. The Trapezius insertion into the shoulder was 80 mm from the AC joiunt ans Middle belly of the Trapazius was 145 mm from the AC joint. Three indentation tests were completed at the different locations.

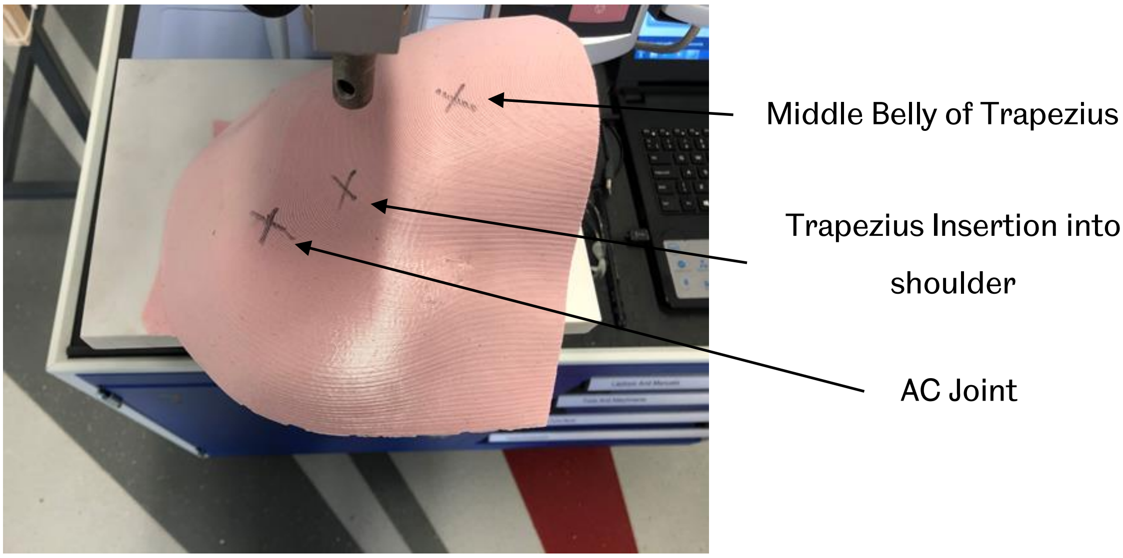


Figure 5.32 – Compression testing set up with loading locations marked.

A stress-displacement trace is pictured in Figure 5.33 with the simplified surrogate overlayed. The load response at the AC joint is much stiffer than the Trapezius insertion into the shoulder and the middle belly of the Trapezius. This is to be expected due to the increased amount of soft tissue in these regions. The median stress-displacement trace for each location was used.

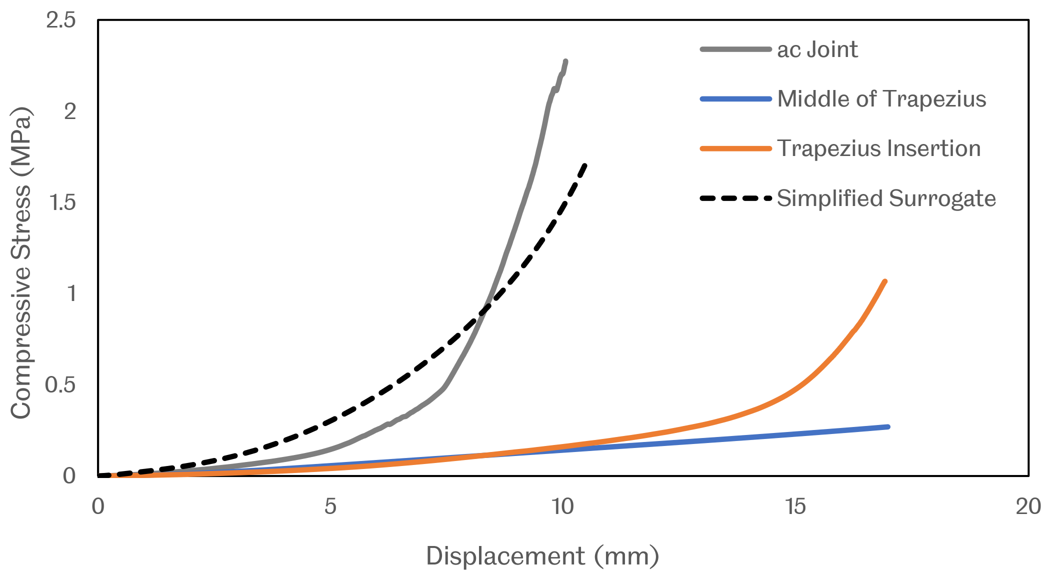


Figure 5.33 – stress-displacement trace at each location (simplified surrogate overlayed).

When comparing the simplified surrogate stress-displacement trace a more comparable trace can be seen with the AC joint than the Trapezius insertion into the shoulder and the middle belly of the Trapezius. This could be because of two reasons, the greater amount of soft tissue in that region and the simplified surrogate having a chamois leather skin layer which will stiffen it at smaller displacements.

There was a good consistency of response when repeating loading cycles. This is illustrated in 5.34 where each stress-displacement trace is similar. The only outlier is trace 1 in Figure 5.34a where the response is stiffer. This may be due to a slightly different location of loading, meaning the indenter came into contact with the AC joint at a smaller displacement. This shows the variability of human body structures, especially when compared with 5.34c where the indenter would only have contacted the silicone elastomer.

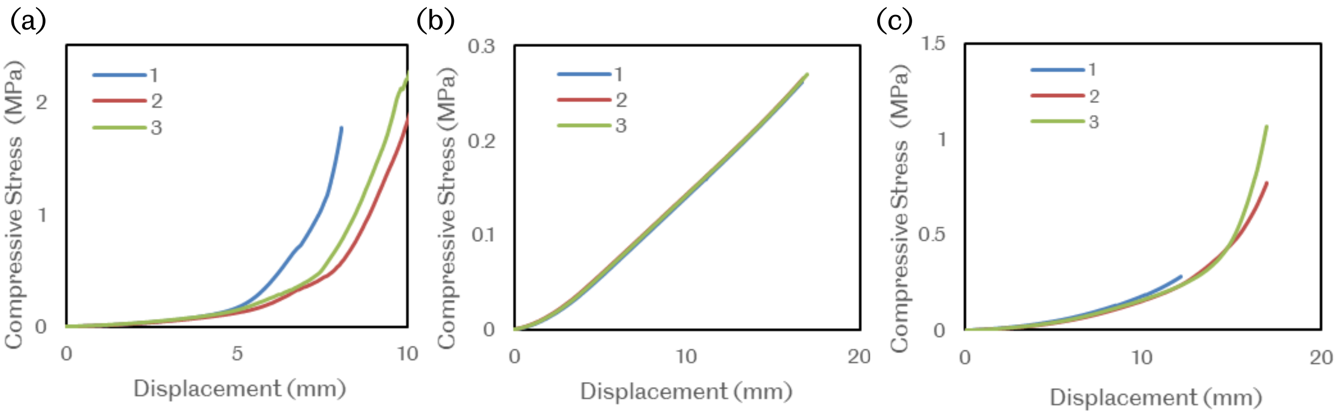


Figure 5.34 - stress-displacement trace at (a) AC joint, (b) Trapezius insertion into the shoulder (c)The middle belly of the Trapezius.

5.5.5 Discussion

Quality of Anatomical Shoulder Surrogate

The anatomical shoulder surrogate aimed to design and fabricate a two-layer surrogate using representative anatomical geometries, for use in the impact testing of shoulder padding. Some anatomical simplifications to internal geometries were made, however, it is intended that the surrogate's dynamic response to impact will more closely match what is seen in the human equivalent when compared to what has been presented to date. When compared to both a steel cylinder surrogate used in current Regulation 12 test procedures and the simplified surrogate developed in §5.4 the anatomical surrogate provides a far superior representation of the external geometries in the human shoulder. These geometries will vastly affect the surrogate's response to impact, so ensuring these are accurate is crucial. As well as this, the surrogate internal bone and muscle geometries represent the human. A large amount of the shoulder is made up of skeletal components and muscle tissue. Ensuring these geometries are accurate is crucial so that the surrogate's dynamic response at different shoulder locations can be explored.

The anatomical geometries of the surrogate were determined through both 3D and Ultrasound scanning of an appropriate data set of rugby players, which links to the surrogate's application to be used in the impact testing of rugby shoulder padding. No other research has used an anatomical dataset that matches its application. Payne [210] used a data set from Visible Human Project (VHP) [213] and outlined that this was not representative of an athletic population. Understanding how these geometries differ between populations would be beneficial for future research.

The ease of surrogate fabrication can be praised. After the mould and skeletal component were produced, a one-stage process of setting a silicone muscle layer was followed. This will facilitate the replication of further surrogates in the future. However, this could become more complexed if a skin or fat layer were to be included.

The surrogate used 3D printing technologies to develop a shoulder with accurate bone geometries. This can be commended due to the complexity of the skeletal components in the shoulder. This two-layer approach where CAD models of the external and internal geometries of a human can be applied to any other body part where a soft tissue region surrounds a skeletal component.

Anatomical Shoulder Surrogate Improvements

The shoulder surrogate only embodies a two-layer approach due to ease of fabrication and the fact that the shoulder is mainly made up of bone and muscle. However, because of this, some biofidelity is lost. This is mostly due to an absent skin layer that will affect the surrogate's response to impact at its external geometries due to its elasticity. The absent skin layer also means Cut, Laceration and Abrasion injuries cannot be explored. This becomes an issue when considering World Rugby's™ RQ 3 as padding's ability to protect from these injuries cannot be explored. Moulding a chamois skin layer to the surrogate's complex geometries would prove difficult, therefore the addition of a silicone skin layer could prove a more appropriate alternative.

An adipose tissue layer was not considered, mainly due to there being a small amount of it in the shoulder region of humans. Adipose tissue does, however, have impact attenuative properties, coupled with this, humans with a greater fat mass will in turn have a thicker adipose tissue layer between their skin and muscle in the shoulder. A four-layer surrogate with a skin and fat layer could be considered, however, fabrication would become more difficult. Other anatomies like vascular tissue or cartilage were not considered, reducing the surrogate's biofidelity. The addition of these tissues in future surrogates could be considered, however, this would cause additional complexities in fabrication, and whether they affect impact response is questionable.

The muscle layer of the shoulder surrogate represents relaxed muscle tissue. The effect of muscle contraction on the mechanical response of muscle to impact was therefore not considered. Many biomedical studies have investigated the effect of muscle contraction on its mechanical characteristics [216-218]. However, current literature provides no conclusive data on how levels of contraction can change the response of the tissue. Tackle biomechanics research shows the trapezius muscle is contracted when making a rugby tackle [61]. When considering the application of this shoulder surrogate, further research is required in the development of contracted muscle simulants.

Because of the surrogate design, it featured a Clavicle support that acted as the non-existent sternum. How this may affect its response to impact is unknown. Extending the overall surrogate medially to embody the sternum and part of the neck could produce a more biofidelic response to impact both in the trapezius and its underlying structures like the clavicle. However, this would increase the complexity and size of the surrogate, increasing its cost and decreasing its ease of fabrication and replication.

5.6 Summary

The chapter details the development of custom-made silicones tailored to match the response of human soft tissue. And the development of both a simplified and anatomical human shoulder surrogate. It is important to assess the pros and cons of each surrogate, why each surrogate was developed, and highlight the plans for impact testing of each surrogate. Table 5.13 highlights each surrogates' pros and cons.

Table 5.13 – Pros and cons of each surrogate developed.

	Pros	Cons
Simplified Shoulder Surrogate	*Uniform design making it repeatable, enabling test house use. *Skin layer incorporated for the assessment of Cut, Laceration, and Abrasion injuries. *Design enables easy fabrication so that many surrogates can be produced in a repeatable manner. *Is cost-effective when compared to other sports impact surrogates (§5.4.5).	*Many geometrical simplifications have been made, potentially altering the shoulders 'true' response to impacts. *The surrogate has not been validated using actual load data from humans.
Anatomical Shoulder Surrogate	*Geometrically accurate external and internal anatomies have been modelled. *The surrogate enables the ability to assess for load propagation and distribution of impacts on differing shoulder locations. *The design of the surrogate and mould makes it easy to repeat fabrication when needed.	*The surrogate lacks a skin layer meaning Cut, Laceration and Abrasion injuries can't be assessed. *The surrogate has not been validated using actual load data from humans.

Both the simplified and anatomical shoulder surrogates have been used in impact testing for the assessment of padded clothing (§6). However, they have different uses in relation to the assessment of padded clothing, Table 5.14 explains this providing rationale as to why both surrogates were developed.

Table 5.14 – Intended use in impact testing of each surrogate.

Surrogate	Intended use in impact testing
Simplified Shoulder Surrogate	*The surrogate embodies a skin layer, this can be used to assess padded clothing's ability to protect from Cut, Laceration, and Abrasion injuries both from blunt force (i.e. a player's knee) or from a player's stud. *The uniform and repeatable design of the surrogate means it can be used to assesses the force attenuation properties of varying designs of padded rugby clothing add differing impact energies. An accurate and reliable assessment can then be made.
Anatomical Shoulder Surrogate	*The anatomical design of the surrogate means the load propagation and distribution of an impact can be assessed at different anatomical landmarks of the shoulder. It is expected that an impact on the trapezius will display different properties to an impact on the AC joint. *The anatomical design also means the performance of padded clothing on differing impact locations on the shoulder can be assessed.

CHAPTER 6 - THE ASSESSMENT OF PADDED CLOTHING'S PROTECTIVE CAPABILITIES

6.1 Chapter Overview

This chapter presents all experimental impact testing procedures used to determine the ability of padded rugby clothing to protect against common impact injuries that take place on a rugby pitch. These testing procedures have also been used to determine the impact response of the shoulder surrogates developed in chapter 5. Impact tests have been carried out to replicate Laceration, Cuts, and Abrasion injuries caused both by blunt force (bony parts of the human) or player's studs. Impact tests were conducted recording accelerations, peak force, and displacement using an accelerometer, load cell, and high-speed video. The results of this testing have also been used to guide the recommendations made to World Rugby™ regarding the performance requirements of padded rugby clothing (shoulder padding).

Original plans were to use a bespoke impact drop tower developed at UoS to assess padded clothing's performance. However, due to various factors including, time constraints, costing and the Covid-19 pandemic this did not get completed. Designs of this impact drop rig can be seen in Appendix C. Because of this, the impact drop rig used in §3.3 that replicated Regulation 12 test standards were modified to incorporate the testing protocols presented in the chapter. This drop rig is detailed in 6.2.

The impact testing procedures for the current Regulation 12 – padded clothing were replicated and reviewed in §3.3. The key conclusions made from this regarding the impact test were that:

1. The impact test setup does not represent a rugby impact.
2. The impact test protocols make the repeatability of testing problematic.
3. The performance requirements have no documented research to back them up.

The work in this chapter looks to use experimental testing procedures to develop a greater understanding of these conclusions and in turn guide new impact testing protocols for padded clothing Rugby Union. Throughout the chapter, the term 'severity' is used to describe the impacts. A more severe impact will be caused as one or more of the striking bodies is more rigid and therefore the contact time to peak impact force is shorter.

6.2 Drop Rig used for the Assessment of Padded Clothing

It is important to introduce and explain the drop rig used for all the impact testing completed in this thesis. This section highlights its design and the instrumentation used as well as how impact testing setups could be changed. The drop rig was located at MMU in a fixed permanent position. It incorporated a uniaxial design with a drop weight on steel rail guides. When the magnetic switch was released the drop weight accelerated towards the intended impact location. The mass of the drop weight could be altered and the impactor interchangeable. An accelerometer could be attached to the impactor to measure impact acceleration. The anvil (impacted object) could also be interchanged to allow for different impact surrogates to be fitted. Load cells were placed under the anvil so impact force could be measured. A high-speed camera was also utilised to allow for 2D photogrammetry. Two separate setups can be seen in Figures 6.1a and 6.1b highlighting their interchangeability.

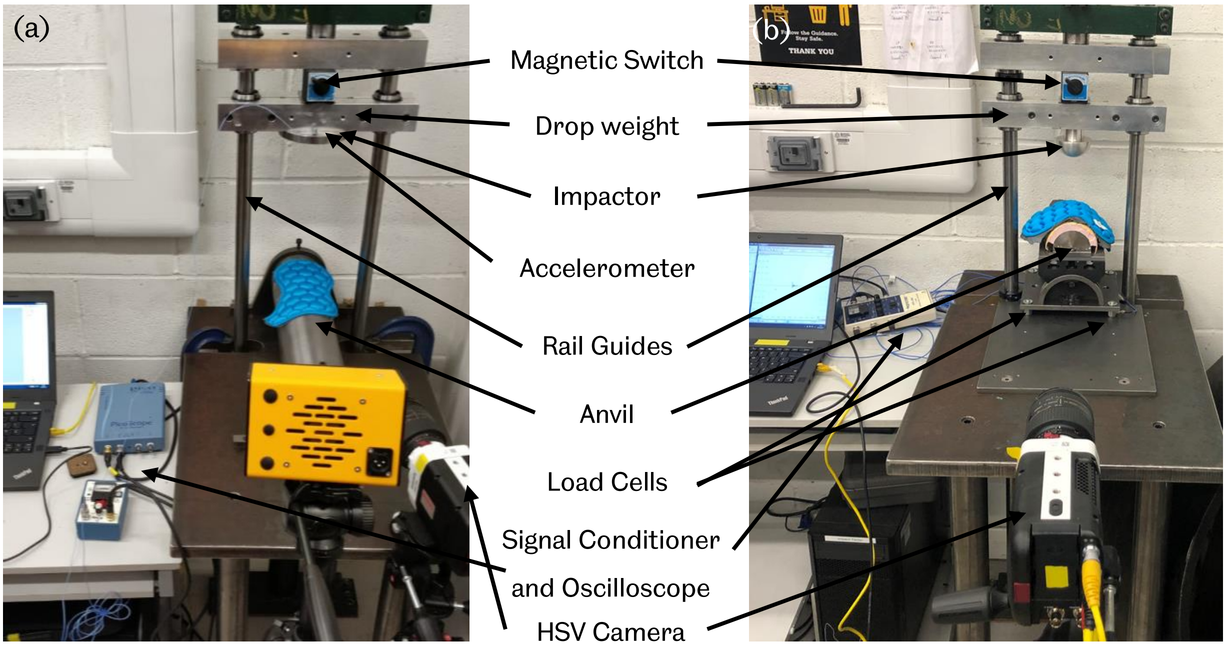


Figure 6.1 – (a) Drop rig set up (as per Regulation 12) with accelerometer fitted, (b) Drop rig set up with simplified shoulder surrogate integrated and load cells fitted.

Instrumentation

The instrumentation used in the drop rig was key to assessing the performance of padded clothing, it is listed below. All filtering was completed automatically by the signal conditioner, therefore no manual filtering was needed.

- Load Cells (208C05-Force Sensor, 22.34 kN measurement range, PCB®, Piezotronics) – Four load cells were fitted under the anvil sampling at 20 kHz were connected to an oscilloscope (PicoScope®, Version 6, Pico Technology) via two 3-Channel ICP sensor signal conditioners (480B21, PCB®) to record impact force. This was pre-calibrated by the manufacturer.
- Accelerometer (352B01-ICP-Accelerometer, 5000 g measurement range, PCB®, Piezotronics) – A uniaxial accelerometer could be fitted to the impactor sampling at 20 kHz and connected to an oscilloscope software PicoScope® (Version 6, Pico Technology) via an ICP® sensor signal conditioner (480B21, PCB®), to enable temporal acceleration to be obtained throughout impact. This was pre-calibrated by the manufacturer.
- High-Speed Video Camera (Phantom Miro R111, Vision Research, USA) – A high-speed video camera with a zoom lens (Nikon AF Nikkor 24-85mm 1:2.8-4 D, Nikon Corporation, Japan) could be used to calculate key parameters including impact velocity, contact time and displacement. High-speed video clips could also be used as an important tool in highlighting and describing the type of impacts seen.

Calculation of Impact Velocity Energy and Displacement

Although impact force and acceleration were measured using the instrumentation equipped on the drop rig. Impact velocity (*equation 6.1*) and impact energy (*equation 6.2*) could be theoretically calculated. Whereby g = acceleration due to gravity, h = drop height, and v = velocity, frictional losses in velocity were discounted. This was because when comparing impact velocities (Regulation 12 setup, 5 kg mass, three repeat impacts at 0.1, 0.2, 0.3, and 0.4 m) calculated from HSV footage with theoretical values calculated from equation 6.1, the mean absolute error was only 1.3 %. Displacement (d) in mm was calculated from high-speed camera stills using a known distance in the still by positioning a ruler in the frame and calculating the distance in pixels (P). From this, pixels per mm ($PPmm$) could be calculated. Equation 6.3 could then be used to calculate the displacement in mm.

$$V = \sqrt{2gh} \tag{6.1}$$

$$KE = \frac{1}{2}mv^2 \tag{6.2}$$

$$d = \frac{P}{PPmm} \tag{6.3}$$

6.2.1 Comparison of Load Cell and Accelerometer Data

It was suggested in 3.3.2 that accelerometer data could potentially cause issues with noise and the use of a load cell would provide cleaner more accurate data. It was suggested a load cell was used for the rest of the impact testing, however, providing a rationale for this was crucial.

To do this the drop rig was set up as per regulation 12 (Figure 6.1a). The load cell and accelerometer were instrumented to the drop rig. Impact tests were completed using a mass of 5 kg (flat face, Ø 130 mm) from 0.3 m onto the rigid cylindrical anvil and a commercial padding sample. Five repeats were performed with the median max voltage being displayed in figure 6.2. Figure 6.2 shows the response of the accelerometer with the load cell's response overlayed. It is clear that the data acquired from the accelerometer produces a large amount of noise when compared with the load cell. Figure 6.2 also suggests the accelerometer could be overestimating the resultant peak force. It is important that an accurate peak force value is attained when using the drop rig for the impact testing of padded rugby clothing. With this in mind, the use of a load cell to acquire force data was used throughout the rest of the impact testing described in the thesis.

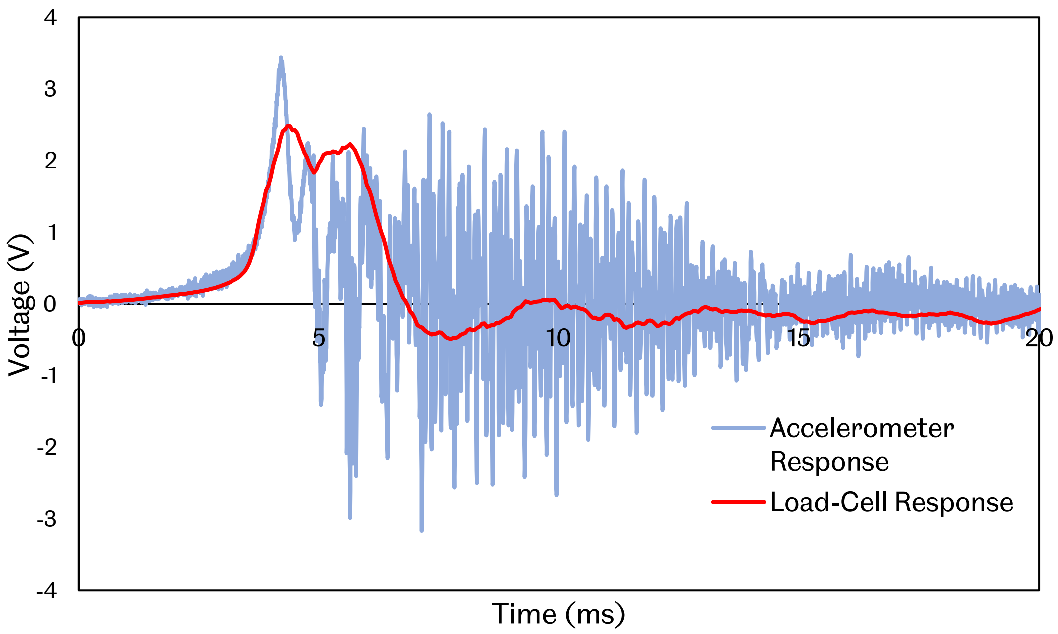


Figure 6.2 – Voltage response of accelerometer and load cell.

6.3 Assessment of Padding using Rigid Impacts

6.3.1 Introduction

When critiquing the current Regulation 12 procedures (§3.3). It was clear that the peak forces in the impact test were far greater than what is seen in an impact in Rugby Union. Coupled with this, there is no knowledge of how much impact force padded clothing is attenuating, this is a key characteristic with regard to the performance of the padding. This section therefore aims to:

1. Assess how much impact force current padded rugby clothing is attenuating.
2. Assess how differing the impact energy of the Regulation 12 impact test can vary the peak impact forces outputted.

With these aims in mind, this section presents the impact testing completed using the Regulation 12 test procedures at differing impact energies both with and without the addition of both the control material (Plastazote) and commercial padding samples.

6.3.2 Impact Testing with Rigid Striker and Anvil

This section presents the impact testing completed using the Regulation 12 impact test protocols, without the addition of padding samples. Therefore, the steel striker would impact with the steel anvil causing an almost rigid impact and the peak force recorded. Due to load cell capacities, this was done at low impact energies, the results were then extrapolated as well as compared with peak force values from an FE model developed by PhD A. These results could then be used to calculate the amount of force attenuated when padding was added in the same impact test setup.

Test Methodology

Impact tests were completed at impact energies of 0.5, 1, 1.5, 2, 2.5 J (rounded to one decimal place) using a 5 kg mass (flat face, Ø 130 mm) dropped from 0.01, 0.02, 0.03, 0.04 and 0.05 m (Figure 6.2) This was calculated using equation 6.2. The anvil (horizontal steel cylinder, Ø 115 mm) was fixed on four load cells (208C05-Force Sensor, PCB Piezotronics) that had a sampling rate of 20 kHz and were connected to an oscilloscope (PicoScope®, Version 6, Pico Technology) via two 3-Channel ICP® sensor signal conditioners (480B21, PCB®) to record impact force. The impact was completed three times at each energy with at least one minute between each impact at room temperature (20 ± 2°C). The voltage readings from each load cell were converted to force using the calibration factors (range: 0.2214 to 0.2399 mVN^{-1}) provided by the manufacturer. Force–time traces were outputted.

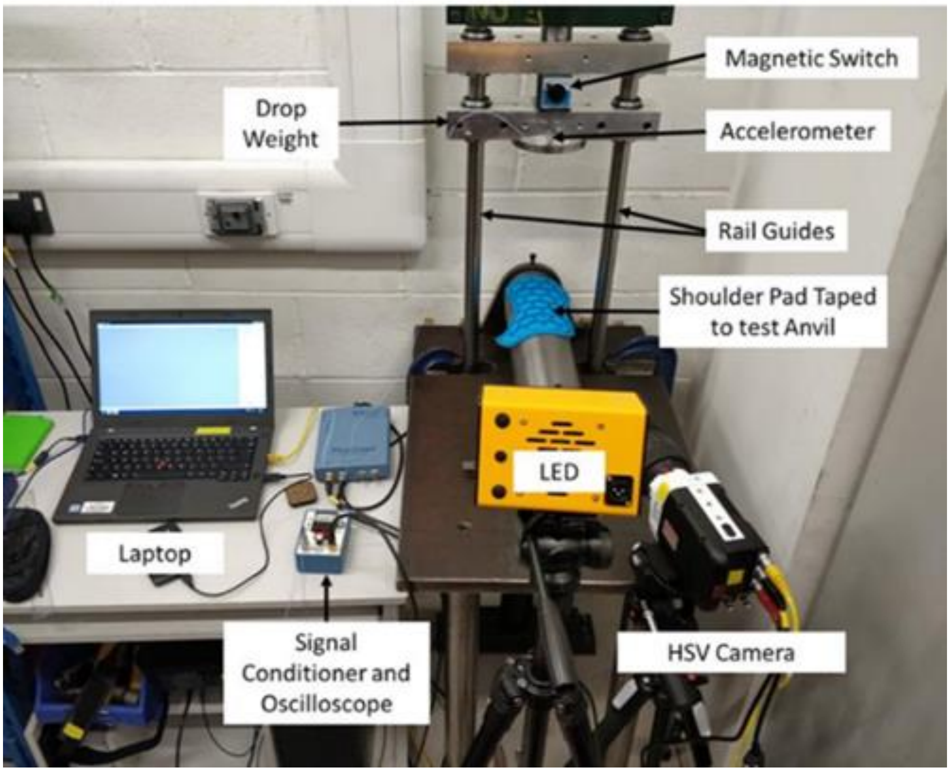


Figure 6.3 – Drop rig setup (as per Regulation 12) used for rigid body impact testing (note: padding sample in the image was not included in testing).

Results

Peak force values of the corresponding impact energies are shown in Table 6.1. As well as this, peak force values at impact energies of 4.9, 9.8, and 14.7 J are displayed, this has been calculated using extrapolation from the linear trend line seen in Figure 6.4. The peak force values calculated using an FE model developed by PhD A of the World Rugby Regulation 12 project (§1.30 for a steel on steel impact are also displayed.

Table 6.1 – Mean peak force measurements at differing energies using rigid body impact Regulation 12 set up, Mean ± SD (Three impacts).

Drop Height (M)	0.01	0.02	0.03	0.04	0.05	0.1	0.2	0.3
Energy (J)	0.5	1.0	1.5	2.0	2.5	4.9	9.8	14.7
Peak Force (N)	2695 ± 20	3635 ± 18	4627 ± 7	5508 ± 45	6180 ± 56	10719*	19563*	28407*
FE Peak Force (N)	2645	3818	4594	5466	6097	9810	19620	29430

* 4.9 J, 9.8 J, and 14.7 J have been extrapolated from the linear trend line.

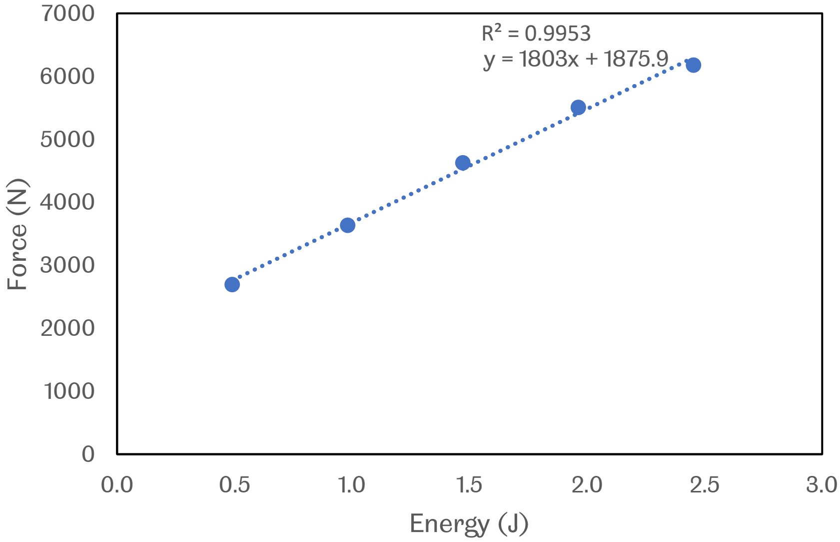


Figure 6.4 – Peak force plot using mean experimental data with linear trend line.

Forces extrapolated from the experimental values and FE model forces at energies of 4.9, 9.8, and 14.7 J are presented in Table 6.2. The FE results offer validation for the extrapolated peak force values. However, it should be noted that FEA has assumptions, with inputs from experimental data so should not be used as a gold standard for validation. The percentage difference from the FE results has also been calculated with the largest being 8.5 %. Extrapolated peak force values would therefore be used to estimate the peak force of a steel on steel impact at these energies. These impact energies were chosen because 14.7 J is the Regulation 12 standard. However, as mentioned in §3.3 peak forces in the impact test are far greater than a rugby impact. therefore, peak forces at 9.8 and 4.9 J were also estimated. In theory, using the linear trend line, the rigid body peak impact force at any energy could be calculated.

Table 6.2 – Extrapolated peak force compared with FE model.

Energy (J)	4.9	9.8	14.7
Extrapolated Peak Force (N)	10719	19563	28407
FE Peak Force (N)	9810	19620	29430
% Difference	8.5	0.3	3.6

6.3.3 Impact Testing with Rigid Striker/ Anvil and Padded Clothing

This section presents impact testing completed using the Regulation 12 impact test protocols, with the addition of manufacturers' padding samples at a range of impact energies. This was in addition to the testing presented in §3.3 at 14.7 J. Using the results in §6.3.2 the force attenuation of the padding could be calculated, this is a key characteristic of the protective properties of padding. Manufacturers' samples and the control material used in §3.3 were impacted at 9.8 and 4.9 J. The same test methodology highlighted in §6.3.2 was used; however, the padding was fixed to the steel anvil. Peak force was divided by the product of the drop mass (5 kg) and the gravitational constant (9.81 $m.s^{-2}$) to give peak acceleration during impact, in g. Three impacts were completed as per Regulation 12, a new impact site on the padding was selected for each impact.

Results

Peak force and acceleration values at corresponding energies for each sample are shown in Table 6.3. The control material (Plastazote) was chosen because its impact attenuation capabilities sit near the minimum pass mark for the current Regulation 12 impact test (150g). The mean peak force exhibited by rugby padding increases at a similar percentage when compared with impact energy (200 % increase from 4.9 to 14.7 J, 189 % increase from 3714 to 10689 N). The control material was left out of mean force calculations as it was not an approved padding material.

Table 6.3 - Peak impact force and acceleration at 4.9, 9.8, and 14.7 J on the rigid cylindrical anvil. Mean ± SD (Three impacts).

	4.9 J		9.8 J		14.7 J	
Sample Number	Peak Force (N)	Peak Acceleration** (g)	Peak Force (N)	Peak Acceleration** (g)	Peak Force (N)	Peak Acceleration** (g)
1	2796 ± 69	57 ± 1.4	7,505 ± 128	153 ± 2.6	10787 ± 59	220 ± 1.2
2	2796 ± 74	57 ± 1.5	7,407 ± 34	151 ± 0.7	10836 ± 34	221 ± 0.7
3	3483 ± 103	71 ± 2.1	7,995 ± 309	163 ± 6.3	11229 ± 54	229 ± 1.1
4	3041 ± 5	62 ± 0.1	7,259 ± 216	148 ± 4.4	10934 ± 142	223 ± 2.9
5	5199 ± 221	106 ± 4.5	8,388 ± 98	171 ± 2.0	11375 ± 64	232 ± 1.3
6	3237 ± 113	66 ± 2.3	7,308 ± 147	149 ± 3.0	10934 ± 69	223 ± 1.4
7	2747 ± 113	56 ± 2.3	6,769 ± 196	138 ± 4.0	10297 ± 255	210 ± 5.2
8	5543 ± 128	113 ± 2.6	8,878 ± 216	181 ± 4.4	10885 ± 113	222 ± 2.3

9	4464 ± 206	91 ± 4.2	8,535 ± 123	174 ± 2.5	10787 ± 216	220 ± 4.4
10	3434 ± 387	70 ± 7.9	7,652 ± 319	156 ± 6.5	10542 ± 368	215 ± 7.5
11	4120 ± 39	84 ± 0.8	7,161 ± 29	146 ± 0.6	8973 ± 64	183 ± 1.3
12c	2158 ± 34	44 ± 0.7	4,660 ± 206	95 ± 4.2	7600 ± 279	155 ± 5.7
Mean*	**3714 ± 132**	**76 ± 3**	**7745 ± 165**	**158 ± 3**	**10689 ± 634**	**218 ± 3**

*control material left out of calculations for mean force
**peak acceleration estimated using F=ma

Figure 6.4 displays force-time traces (4.9, 9.8, 14. 7 J) for both Plastazote and sample number 6 as this had the median peak force of the manufacturers' samples. As expected at higher impact energies, it took a shorter amount of time to reach peak force, and therefore maximum displacement of the striker.

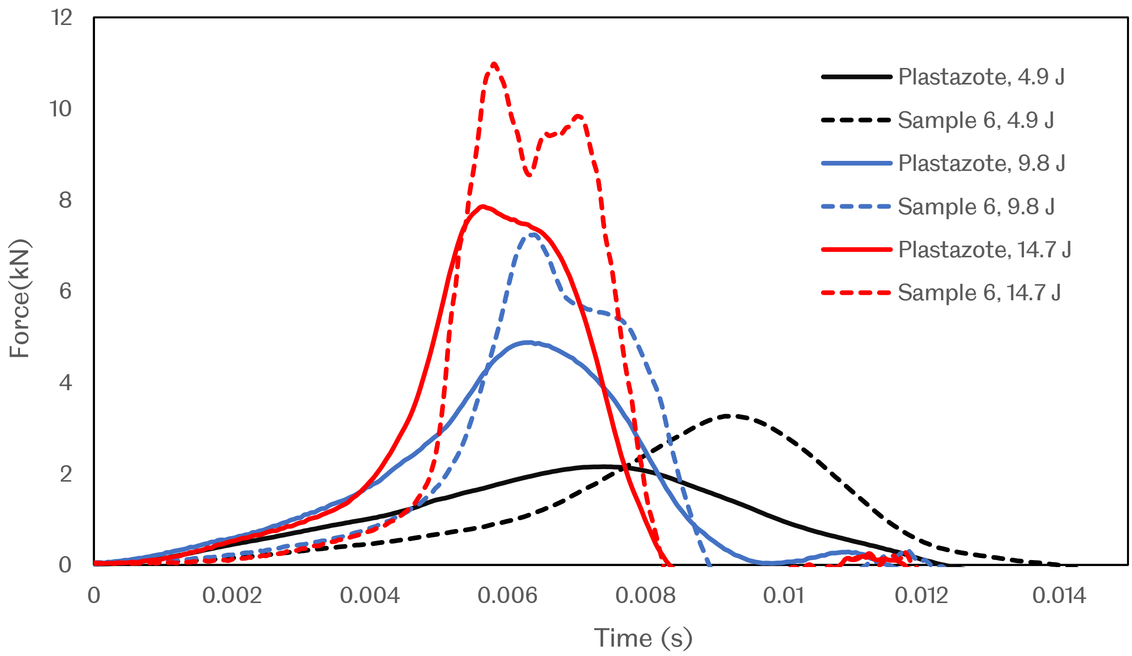


Figure 6.5 – Force-time traces of impacts with Plastazote and samples number 6 at 4.9, 9.8, and 14.7 J.

At 14.7 J there is also a second peak (Figure 6.5), this effect is also starting at 9.8 J. This suggests the padded material had 'bottomed out'. This is where the striker will have contacted the rigid anvil, this is something that could impact the data outputted in a test house. Further exploration of high-speed video confirms this (Figure 6.6). Figure 6.6a shows a capture of sample 6 at maximum displacement, and Figure 6.6b displays a capture of Plastazote at maximum displacement. The striker's displacement from the first contact with the padding was 1.2 mm greater in sample 6. Although this is a small amount, this led to the 'bottoming out' of the sample.

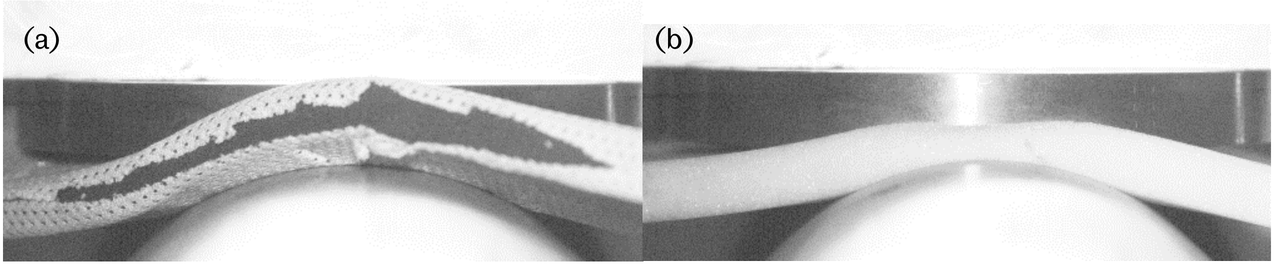


Figure 6.6 – High-speed video capture at the maximum displacement of (a) sample 6, (b) Plastazote.

Table 6.4 compares the mean peak impact force and rigid body peak impact force as presented in §6.3.3 for corresponding impact energies. Using these values, the percentage reduction in force has been calculated. This, therefore, displays the amount of force attenuated by the padding. This ranges from 60.3 – 65.3 % with the mean being 62.7 %.

Table 6.4 – Force attenuation of current manufacturers samples (%).

	4.9 J	9.8 J	14.7 J
Rigid body impact (peak force (N))	10719	19563	28407
Addition of manufacturers samples (mean peak force (N))	3714	7754	10689
Reduction in force from rigid impact (%)	65.3	60.3	62.4

6.3.4 Impact Testing with Flat Rigid Striker and Simplified Surrogate

This section presents the initial impact testing completed that incorporates a more representative anvil to act as a shoulder surrogate. The flat striker used in regulation 12 was kept in order to inform the work carried out in section 6.4 where Laceration, Cuts, and Abrasion injuries were replicated. A similar test set-up, as shown in figure 6.3 was used however the simplified shoulder surrogate acted as an anvil. Illustrated in figure 6.7.

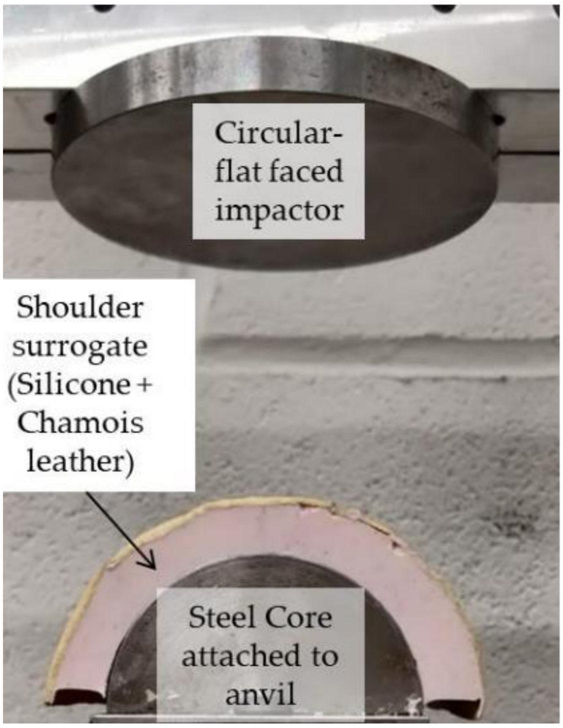


Figure 6.7 – Drop rig set up with flat-faced striker and simplified shoulder surrogate.

The same test methodology highlighted in §6.3.2 was used; however, impacts were completed at 4.9, 9.8 and 14.7 J with no padding. Three impacts were completed as per Regulation 12, a new impact site on the padding was selected for each impact. Peak force was divided by the product of the drop mass (5 kg) and the gravitational constant (9.81 $m.s^{-2}$) to give peak acceleration during impact, in g.

Results

Peak force and acceleration values at corresponding energies for each sample are shown in Table 6.5. When plotting impact energy (x) with max impact force (y), a linear trendline of y = 315.92 x + 986.67 is seen. This trendline can be used to make a direct comparison with the data presented in 6.4.2. Figure 6.8 displays the force-time trace of the median impacts at 4.9, 9.8 and 14.7 J. At higher impact energies, contact time is reduced while peak force is increased.

Table 6.5 - Peak impact force and acceleration at 4.9, 9.8, and 14.7 J on the simplified surrogate. Mean ± SD (Three impacts).

4.9 J		9.8 J		14.7 J	
Peak Force (N)	Peak Acceleration* (g)	Peak Force (N)	Peak Acceleration* (g)	Peak Force (N)	Peak Acceleration* (g)
2457 ± 43	50 ± 0.9	4058 ± 30	83 ± 0.6	5648 ± 152	115 ± 3.0

*peak acceleration estimated using F=ma

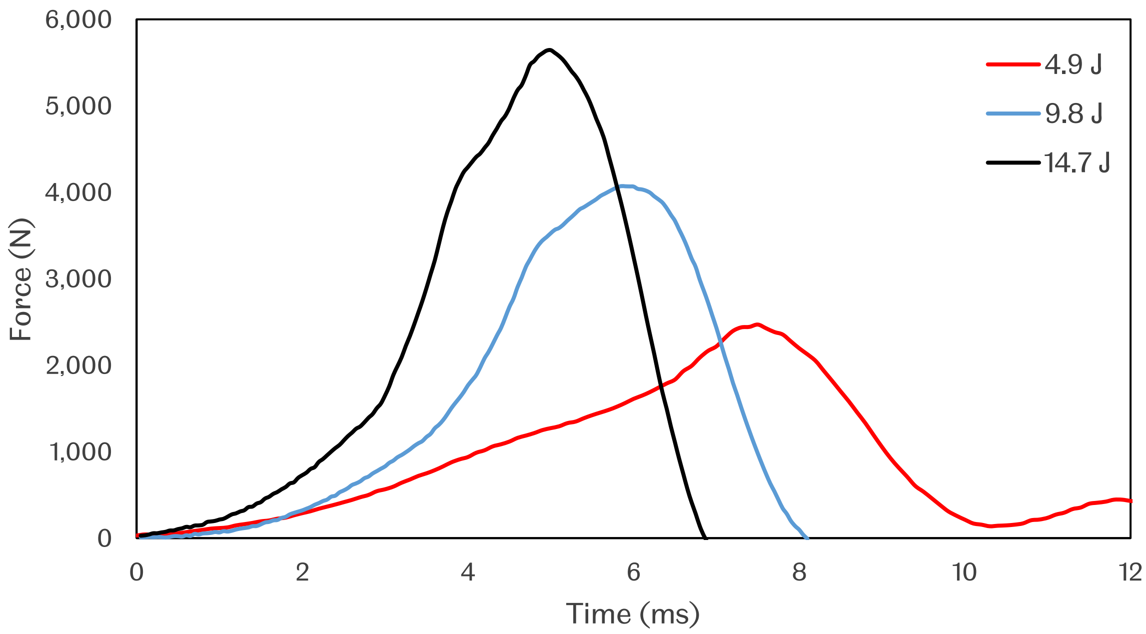


Figure 6.8 – Force-time traces of median impacts at 4.9, 9.8 and 14.7 J

6.3.5 Summary

Sections 6.3.2 and 6.3.3 outline the impact testing completed using a 5 kg steel mass (flat face, Ø 130 mm) and rigid anvil (horizontal steel cylinder, Ø 115 mm) as per Regulation 12. A better understanding of how impact energy can affect the peak force exhibited when the current manufacturers' samples are impacted using the Regulation 12 setup was achieved. Coupled with this, the amount of force attenuated by the current manufacturers' samples at differing impact energies was attained. This being on average 62.7 % of the force. This amount will in turn be a significant characteristic for padding protection, as highlighted in research [104, 219, 220]. However, these are impacts with a rigid striker and anvil, thus they do not represent the human body parts they are embodying (a shoulder and thigh for example). Consequently, this is not replicable of any impacts seen in Rugby Union, therefore, understanding of a more realistic impact situation is needed to assess the efficacy of this approach. The impact testing highlighted in section 6.3.4 will allow for a direct comparison between the use of a flat impactor and a rounder impactor used in section 6.4.

6.4 Assessment of Blunt Force Impact Injuries

6.4.1 Introduction

Section 6.3 successfully assessed padded clothing's impact attenuation abilities. However, it was highlighted that assumptions on padding's ability to protect from specific injuries could not be made. World Rugby™ proposed RQ3 (§1.5) that states:

- Considering that the intention for padded clothing is to continue to protect against Cuts and Abrasions only, devise an updated regulation with testing procedures that permits the latest technology.

To answer this question Laceration, Cuts, and Abrasion injuries needed to be replicated. In the literature review, these injuries could be caused by bony parts of other players or by studs. This section focuses on injuries caused by bony parts of other players, referred to as blunt force injuries.

To replicate Laceration, Cut, and Abrasion injuries caused on the field of play by impacting a bony part (i.e. elbow, knee, shoulder) of another player, a drop rig with a semi hemispherical steel striker was impacted on the shoulder surrogates developed in chapter 5 with and without the addition of padded clothing. This was split into two studies. The simplified shoulder surrogate was used in the first study as it incorporated a skin layer, so tearing parameters could be measured. This also provided a repeatable test, so that padded clothing's protective abilities could be assessed. The second study used the anatomical shoulder surrogate so that differences in impact response at different locations of the shoulder could be assessed with and without padded clothing. These two studies encompassed the following aims and objectives:

1. To determine current padded rugby clothing's ability to protect from Lacerations, Cuts, and Abrasions caused by bony parts of other players.
2. To determine the differences in impact response behaviour at different parts of the human shoulder.

6.4.2 Impact Testing with Rigid Striker and Simplified Shoulder Surrogate

Test Methodology

Impact testing using a semi hemispherical steel striker and simplified shoulder surrogate took place to simulate blunt force Cut/ Laceration impacts. Use of a flat faced striker as in regulation 12 would not simulate a Cut/ Laceration injury so was not used. An incremental process was used to find the impact energy and force the surrogate tore at. This was without padding and with the addition of Plastazote (control) and two manufacturer samples (MS 1 & 2). The experimental test setup is shown in Figure 6.9.

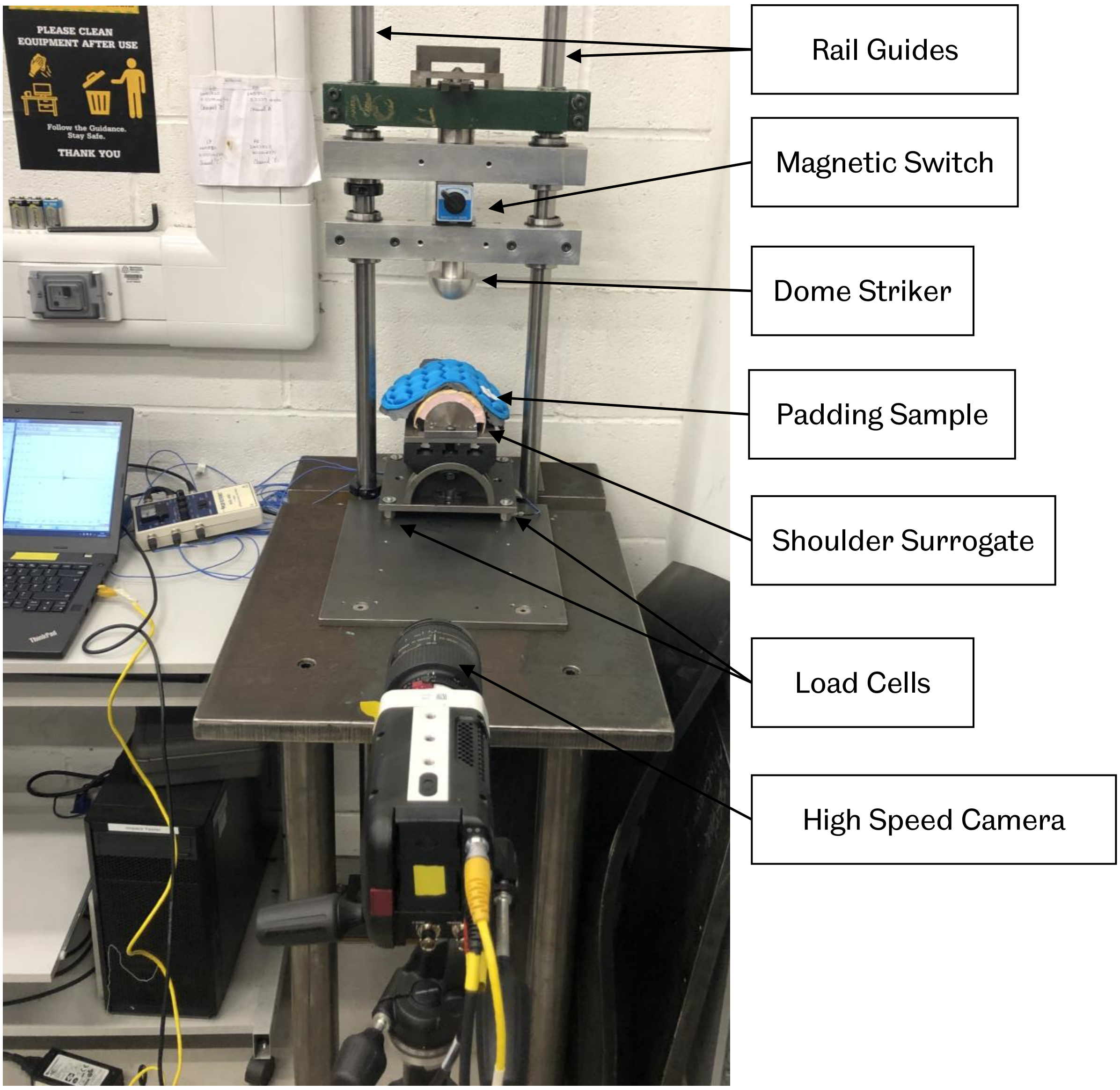


Figure 6.9 – Experimental test set up for replication of blunt force impacts using simplified shoulder surrogate.

The incremental process used was so that the minimum impact energy and force to cause a tear in the surrogate could be found. A tear was defined when the silicone soft tissue layer could be seen through the chamois skin. An initial starting drop height was chosen through experience, and increments were used to limit the number of surrogates needed for testing.

The test procedures are listed below:

1. The semi hemispherical striker (diameter = 75 mm, mass = 3.825 kg) was set to the desired drop height (Table 6.6).
2. The striker was released by switching off the magnet and it impacted with the simplified surrogate.

3. The surrogate was removed from the drop rig and a picture taken, it was then put back into the drop rig, however, a different impacting location was chosen so no degradation from previous impacts was present.
4. The strikers drop height was incrementally increased (+ 5 or 10 cm) and this process repeated until a tear occurred.
5. The same process was repeated but with the addition of Plastazote (control) and two manufacturer samples over the surrogate.
6. Both a force-time trace and high-speed video were recorded to aid the evaluation of results.

Results

High-speed video stills demonstrating the impacts conducted are displayed in Figure 6.10.

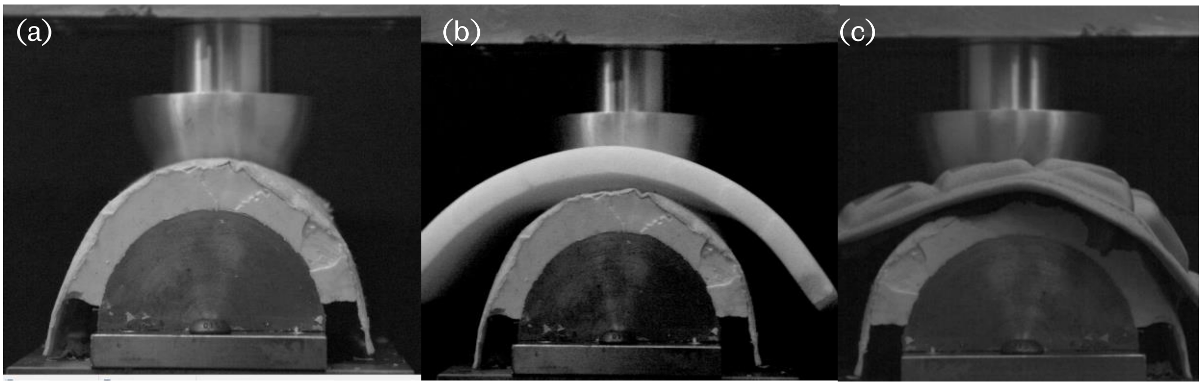


Figure 6.10 – High-speed video stills at max displacement (a) without padding, (b) Plastazote, (c) Padding sample.

The impact energy and peak force of each impact, as well as whether the impact caused a tear is reported in Table 6.6. Further impacts at energies that caused a tear with the addition of padding were also performed without padding.

Table 6.6 – Peak impact force and tear (Y/N) of simplified surrogate impact testing, with and without padding.

Energy (j)		3.8	7.5	9.4	11.3	13.1	16.9	18.8	20.6	22.5	24.4	26.3	28.1	30
Drop Height (M)		0.1	0.2	0.25	0.3	0.35	0.45	0.5	0.55	0.6	0.65	0.7	0.75	0.8
No Padding	Force (N)	1620	4061	5490	6140	7723	-	-	-	12757	-	18288	-	18941
	Tear (Y/N)	N	N	N	N	Y	-	-	-	Y	-	Y	-	Y
Plastazote	Force (N)	-	-	-	-	3097	3864	-	4904	-	6737	-	9243	10663
	Tear (Y/N)	-	-	-	-	N	N	-	N	-	N	-	N	Y
MS 1	Force (N)	-	-	-	-	-	-	7203	-	10331	-	12439	-	13029
	Tear (Y/N)	-	-	-	-	-	-	N	-	N	-	N	-	N
MS 2	Force (N)	-	-	-	-	-	-	6380	-	8149	-	10235	-	10841
	Tear (Y/N)	-	-	-	-	-	-	N	-	N	-	Y	-	Y

A tear is seen at 13.1 J with no padding, an increase of 13.2 J caused a tear with MS 2 added, and 26.9 J with Plastazote and MS 1 added. Images of the torn surrogate can be seen in Figure 6.11. Analysis of the peak force values should be taken with caution as both the padding and the shoulder surrogate will be attenuating a certain amount of this force.

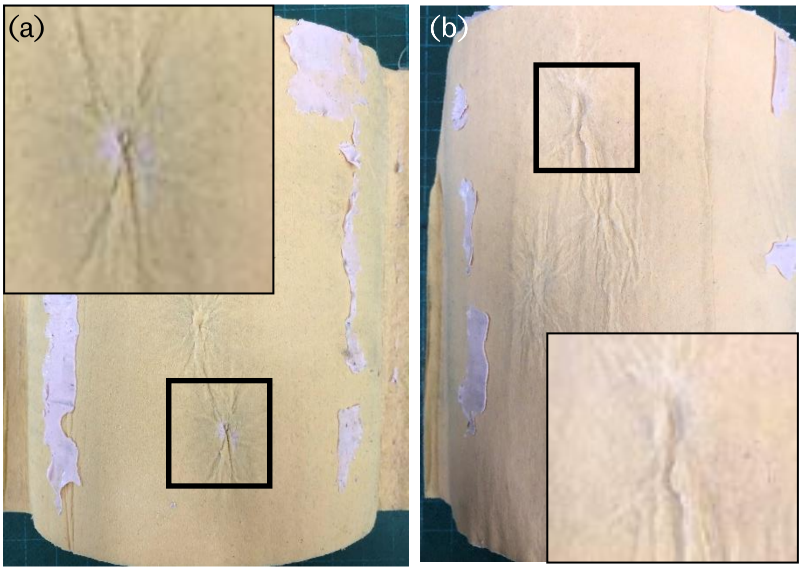


Figure 6.11 – Images of tearing caused by blunt force impacts (a) No padding, (b) Plus Plastazote.

Table 6.7 shows how much extra impact energy and force (as a percentage) is needed to tear the shoulder surrogate when padding was added when compared with the 'No padding' results.

Table 6.7 - Percentage impact energy and force increase (compared to no padding) to exhibit tearing caused by blunt force impacts with the addition of padding.

% Energy increase to tear with padding				% Force increase to tear with padding			
Plastazote	MS 1	MS 2	Mean	Plastazote	MS 1	MS 2	Mean
129	129	100	119 ± 17	80	100	42	57 ± 21

A mean of a 119% increase in impact energy was needed to cause blunt force tearing with the addition of Plastazote, MS 1, and MS 2 compared to no padding. The percentage increase in force is lower (57 %), however, it should be noted that this would be because the padding is attenuating a certain amount of this force.

It was important to gain insights into the amount of impact force both the padding and the shoulder surrogate were attenuating. A similar process was used in §6.3.2 whereby the peak impact force of rigid body impacts was measured at low energies and then extrapolated and compared with FE data. Using this data, the amount of force attenuated by the surrogate and the padding could be measured. To do this, the soft tissue layers of the surrogate were removed, leaving the rigid bone layer. The Rigid impactor was then dropped on the rigid bone layer as seen in Figure 6.12.

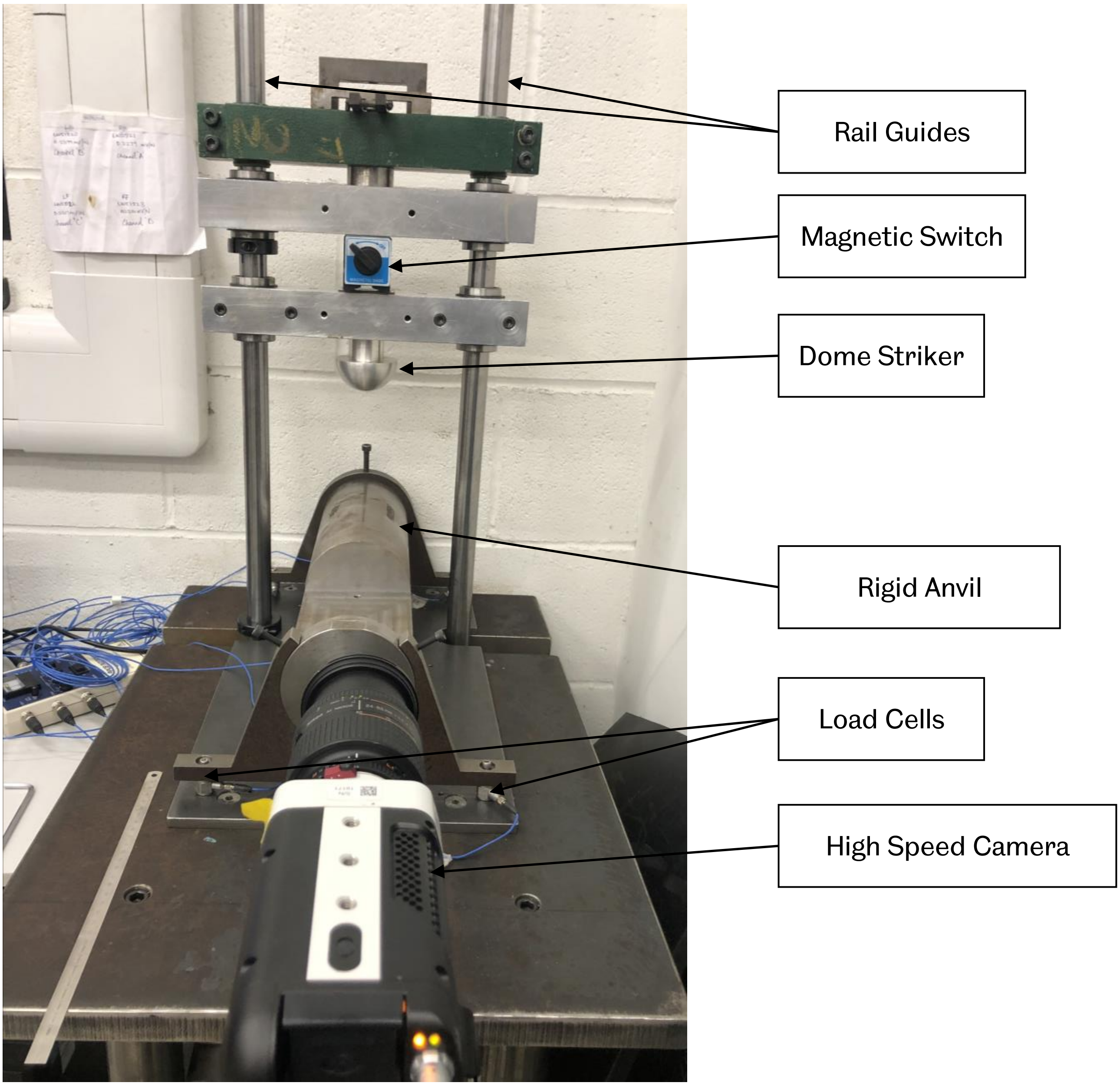


Figure 6.12 – Experimental setup for rigid body impact using a semi-hemispherical striker.

Peak values can be seen in Table 6.8. A linear trend line was calculated from the measured peak forces (Figure 6.13). Peak force (N) at higher impact energies including when surrogate tearing occurred was estimated using extrapolation from the linear trendline (Table 6.9). An FE model was also developed by PhD A. Peak force values and the percentage difference from extrapolated values are displayed. Although the average percentage difference was 20.3 %. As with §6.3.2, the extrapolated values from the measured impacts at lower energies were used for later analysis.

Table 6.8 – Experimental rigid body impact peak force values.

Impact Energy (J)		0.4			1.9			3.8	
Trial	1	2	3	1	2	3	1	2	3
Peak Force (N)	2904	2768	2807	6462	6278	6333	11020	10859	10969
Mean		2826			6357			10949	
SD		70.1			94.6			82.3	

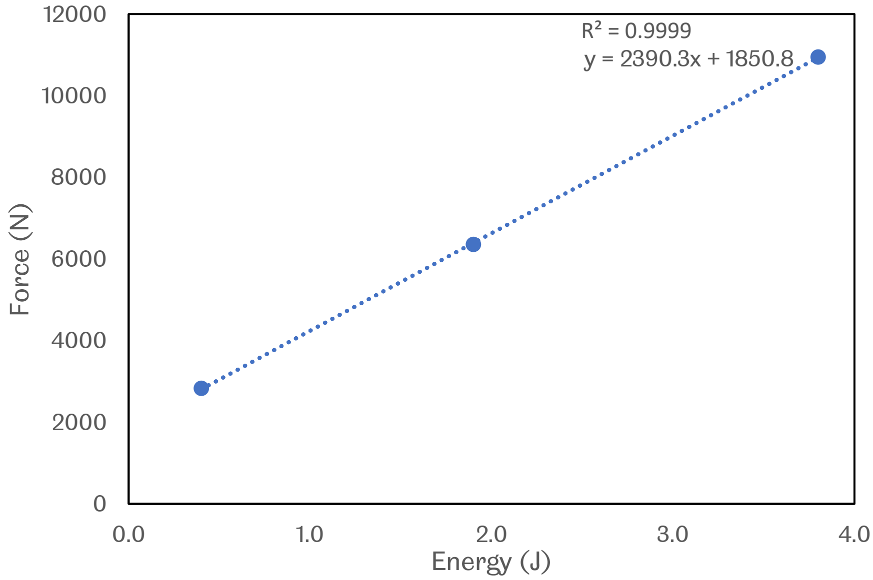


Figure 6.13 – Experimental peak force values with liner trend line.

Table 6.9 – Estimated rigid body impact peak force values.

Impact Energy (J)	7.5	11.3	15.0	18.8	22.5	26.3	30
Extrapolated Peak Force (N)	19789	28758	37728	46697	55666	64635	73560
FE Model Peak Force (N)	15009	22514	30019	37523	45028	52533	60037
% Difference	24.2	21.7	20.4	19.6	19.1	18.7	18.4

Using the results outlined above, the amount of force the simplified shoulder surrogate and the padding is attenuating can be calculated. Once these calculations have been completed, further assessments on padded clothing's ability to protect from Cuts and Lacerations can be made. The peak force of the rigid body impact at certain impact energies has been defined as the **dynamic impact force.** Table 6.10 displays the peak force values exhibited at 13.3 J (when tearing of the chamois skin occurred without padding) and at 30 J (when tearing of the chamois skin occurred with padding added). The percentage reduction in force has also been calculated so that the amount of force the surrogate and the padding are attenuating can be defined. The average of the forces between Plastazote, MS 1, and MS 2 was used.

Table 6.10 – Peak force and force attenuation at tearing with and without padding.

Impact Energy	Impact Test Set-up	Peak Force (N)	Reduction in Force from Rigid-Body Impact (%)
13.3 J (No Padding)	Rigid Body Impact*	28760	-
	Addition of the Surrogate	7720	73
30 J (Padding)	Rigid Body Impact*	73560	-
	Addition of the Surrogate	18940	74
	Addition of the Surrogate and Padding	11150	85

*Peak force has been extrapolated from experimental data.

The results in Table 6.10 show 3 key results that can be used to assess the effectiveness of padding at preventing Cuts, Lacerations, and Abrasions:

1. The surrogate is on average attenuating **73.5%** *((74 + 73) /2)* of the force compared to the rigid anvil.
2. Padding is on average attenuating **41%** *((18940 – 11150) /18940) * 100)* of the impact force with the addition of the surrogate and anvil.
3. An increase of **165%** (44802 N) *(73560 – 28760) / 28760 * 100)* in the dynamic impact force is needed to cause a cut/ tear when padding is added to the surrogate compared to when it is not. If we were to work backward this would be a **61%** *((73560 – 28760) / 73560) * 100)* decrease.

From the results displayed in Table 6.10, there is a range in dynamic impact force where padding is protecting the wearer from a Cut or Laceration that would have been caused without the addition of padding, this is displayed in Figure 6.14 and highlighted orange.

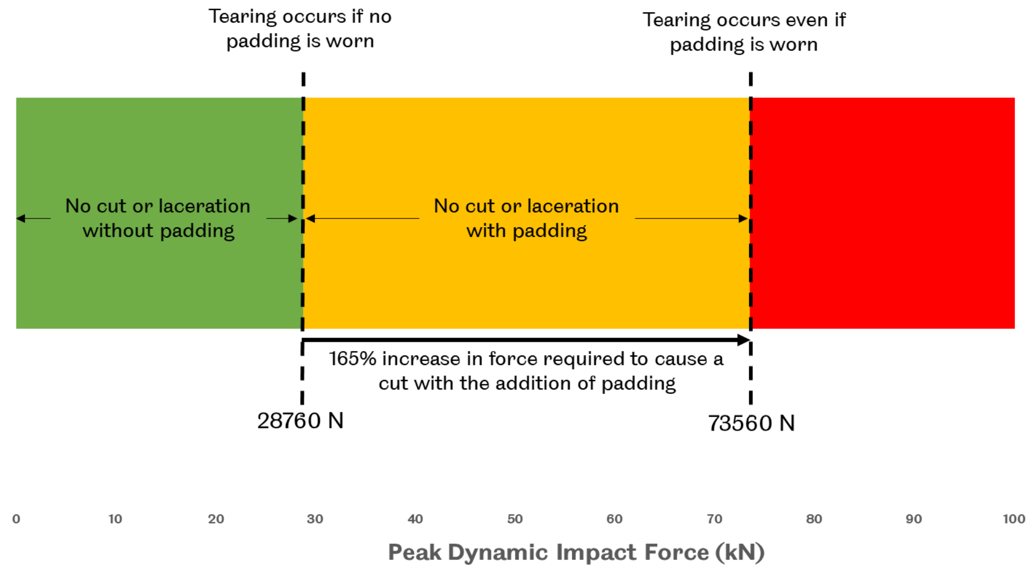


Figure 6.14 – Stacked bar chart of test results.

Repeatability of Testing

To ensure the results of the blunt force testing completed using the simplified surrogate were repeatable. The same testing procedures were completed one week after the original testing. The testing completed was to ensure that:

- Similar force-time traces including peak forces were exhibited.
- The surrogate tore at similar impact energies to the previous testing.

The same testing procedures outlined in §6.4.2 were followed; however, fewer steps were taken in the incremental process to tearing. The results are outlined in Table 6.11. A tear in the surrogate was seen at the same impact energy as the previous testing for all the test conditions (No Padding, Plastazote, MS1, and MS 2). Similar impact forces were also recorded with % differences ranging from 0.2 – 19.8 %.

Table 6.11 – Repeatability testing results.

	Energy (J)	9.4	11.3	13.1	24.4	28.1	30
No Padding	Force (N)	4574	5472	6013	-	-	-
	% Difference	16.7	10.9	16.8	-	-	-
	Tear (Y/N)	N	N	Y	-	-	-
Plastazote	Force (N)	-	-	-	7177	9630	9820
	% Difference	-	-	-	19.8	4.2	7.9
	Tear (Y/N)	-	-	-	N	N	Y
MS 1	Force (N)	-	-	-	10667	12567	13543
	% Difference	-	-	-	3.1	1.0	3.8
	Tear (Y/N)	-	-	-	N	N	Y
MS 2	Force (N)	-	-	-	8132	11005	12564
	% Difference	-	-	-	0.2	7.0	13.7
	Tear (Y/N)	-	-	-	N	Y	Y

* % Difference is calculated from the results displayed in Table 6.6.

These results provide good support that the testing completed with the simplified surrogate is repeatable. This means the results from this can be used to assess the performance of padded rugby clothing in relation to its ability to prevent Cuts, Lacerations, and Abrasions. This is further discussed in §6.4.4.

6.4.3 Impact Testing with Rigid Striker and Anatomical Shoulder Surrogate

The following section outlines the impact testing completed using the anatomical shoulder surrogate developed in §5.5 and rigid dome-shaped impactor. Due to the surrogate not having a skin layer, minimal assessments relating to Cut, Laceration, and Abrasion injuries could be made. However, the impact response at different parts of the shoulder with and without padded rugby clothing could be explored. The test methods and results are outlined below.

Test Methodology

Impact testing using a dome-shaped steel striker and simplified shoulder surrogate took place to simulate blunt force (i.e. elbow, knee, shoulder) impacts. Impacts were completed at three locations on the surrogate as shown in Figure 6.13 both with and without the addition of Plastazote.

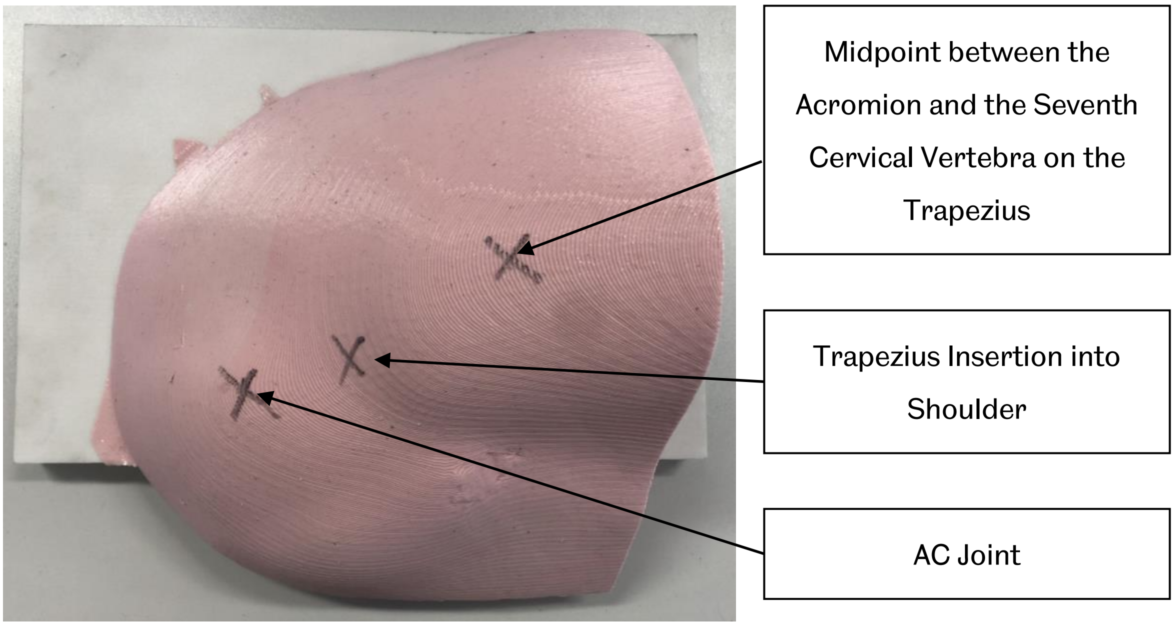


Figure 6.15 Photograph of impacts locations on anatomical shoulder surrogate.

The experimental test setup is displayed in Figure 6.14. This is the same as shown in Figure 6.15 however the simplified surrogate is replaced with the anatomical surrogate.

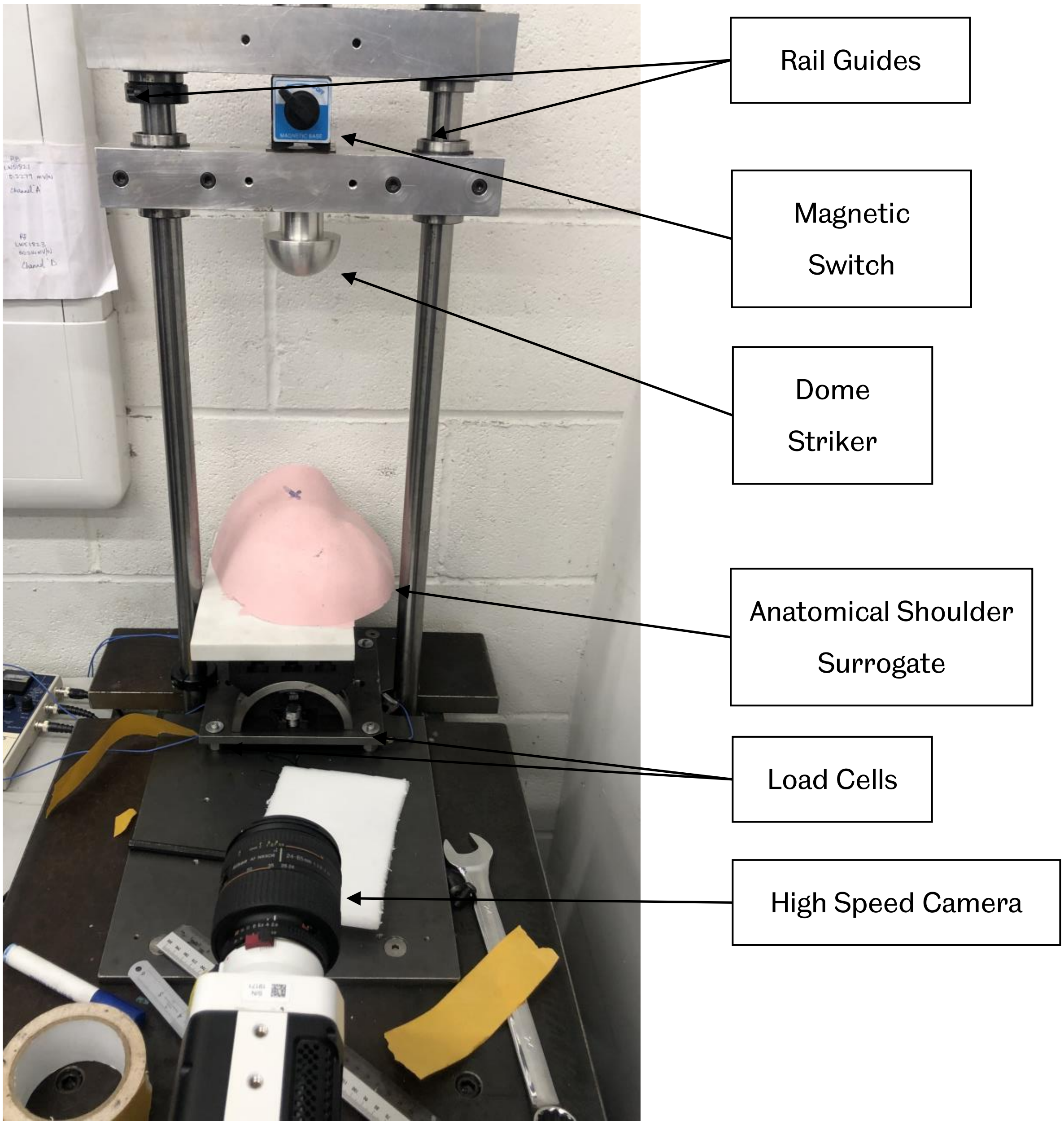


Figure 6.16 Experimental test set up for replication of blunt force impacts using anatomical shoulder surrogate.

Impact energies were chosen based on the energies used in §6.3.4, a range of impact energies mostly in increments of a 10cm drop height were used so that a broader assessment of the impact response of the shoulder surrogate could be made.

The test procedures are listed below:

1. The semi hemispherical striker (diameter =75 mm, mass = 3.825 kg) was set to the desired drop height.

2. The striker was released by switching off the magnet and it impacted with the anatomical shoulder surrogate.
3. The surrogate was removed from the drop rig and a picture taken; it was then put back into the drop rig.
4. The strikers drop height was incrementally increased to the desired impact energy.
5. The same process was repeated with the addition of Plastazote over the surrogate.
6. Both a force-time trace and high-speed video were recorded to aid the evaluation of results.

Results

The surrogate was impacted in various locations with and without padding, peak impact force was attained. High-speed video stills demonstrating the impacts conducted are shown in Figure 6.17.

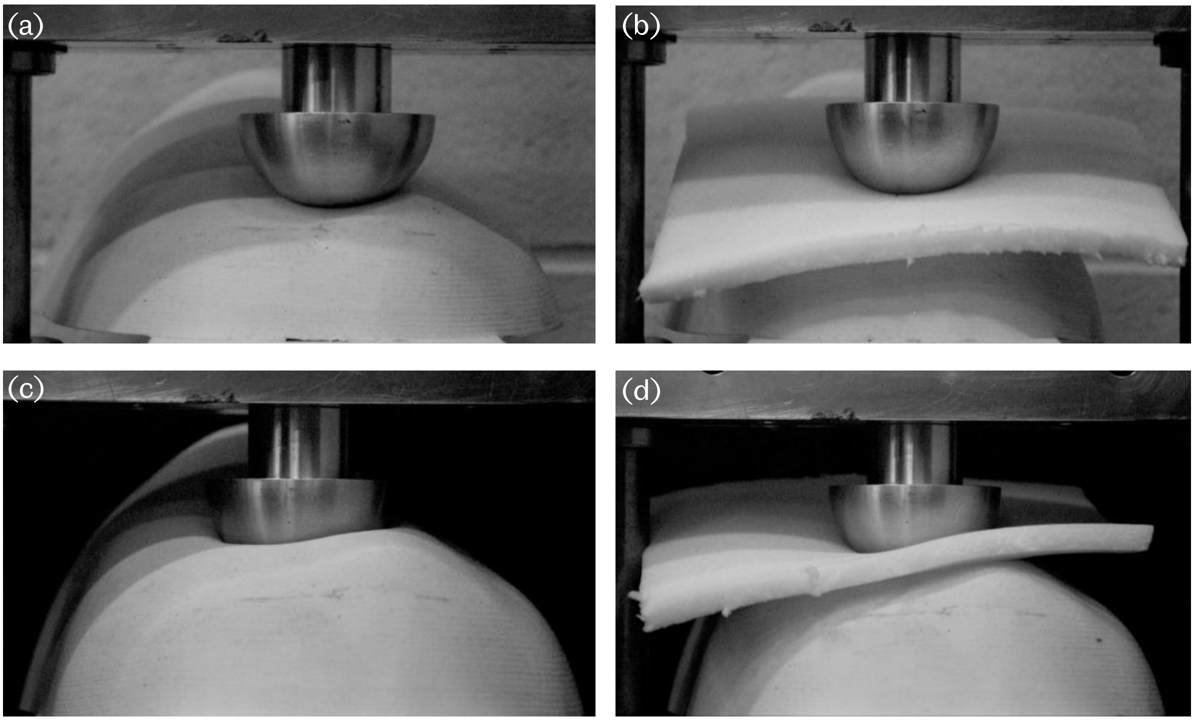


Figure 6.17 – High-speed video stills at max displacement impacting the AC joint (a) without padding, (b) Plastazote and the middle belly of the trapezius (c) without padding, (d) Plastazote.

The impact energy and peak force of each impact, as well as the force attenuation exhibited by the Plastazote have been reported in Table 6.12. The impact on the AC joint is the most severe as there is less soft tissue in that body region, therefore a higher peak force is exhibited compared to the two other impact regions. The impact becomes less severe as the peak force decreases due to the increase in the soft tissue between the striker and rigid bone layer.

An interesting observation is the decrease in force attenuation of the padding as the impact becomes less severe. When impacted on the AC joint padding attenuated a mean force of 67.8 % throughout the energies, this decreases when impacted on the trapezius insertion (26.8 %) and the midpoint between the Acromion and the seventh Cervical Vertebra on the Trapezius (1.8 %). This is further highlighted in Figures 6.18 – 6.20. When impacted on the AC joint, the addition of padding increases the time to peak force, but when impacted on the midpoint between the Acromion and the seventh Cervical Vertebra on the Trapezius the time to peak force is very similar, with comparable force-time traces being displayed.

Table 6.12 – Peak impact force and force attenuation of padding at three impact locations on the anatomical shoulder surrogate (Mean ± SD).

Impact Location	Energy (J)	2.3	4.9	9.8	14.7	18.8
AC Joint	Peak Force (N) – No Padding	1790 ± 355	5483 ± 316	8152 ± 929	-	-
	Peak Force (N) – Padding	438 ± 25	947 ± 67	4454 ± 440	-	-
	Force Attenuation (%)	76	83	45	-	-
Trapezius Insertion into Shoulder	Peak Force (N) – No Padding	-	824 ± 5	1383 ± 59	2555 ± 173	-
	Peak Force (N) – Padding	-	622 ± 23	1023 ± 55	1794 ± 228	-
	Force Attenuation (%)	-	25	26	30	-
Middle Belly of Trapezius	Peak Force (N) – No Padding	-	496 ± 5	813 ± 3	1065 ± 9	1157 ± 25
	Peak Force (N) – Padding	-	497 ± 9	778 ± 9	1034 ± 30	1090 ± 15
	Force Attenuation (%)	-	-0.2	4	3	6

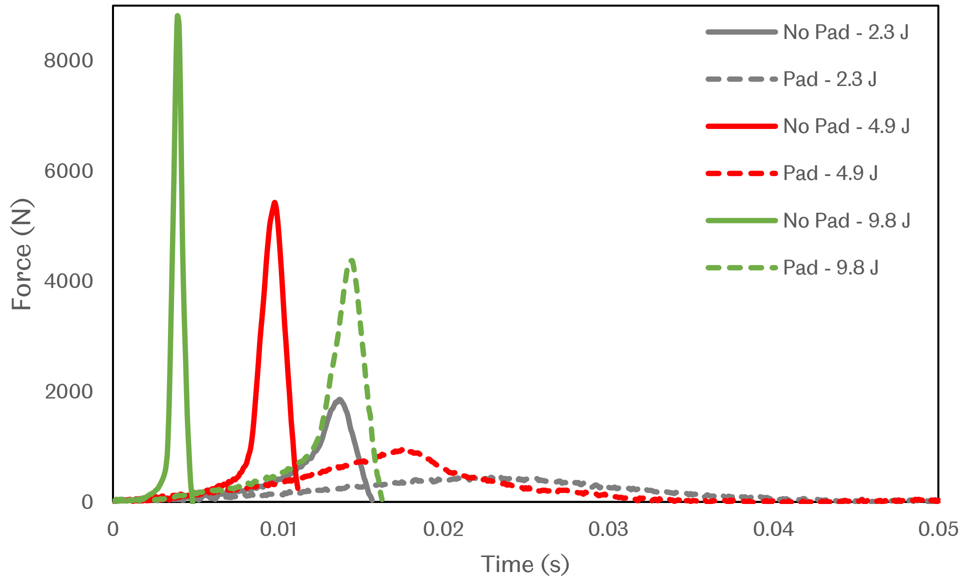


Figure 6.18 – Force-time trace at differing impact energies with and without padding impacting the AC joint.

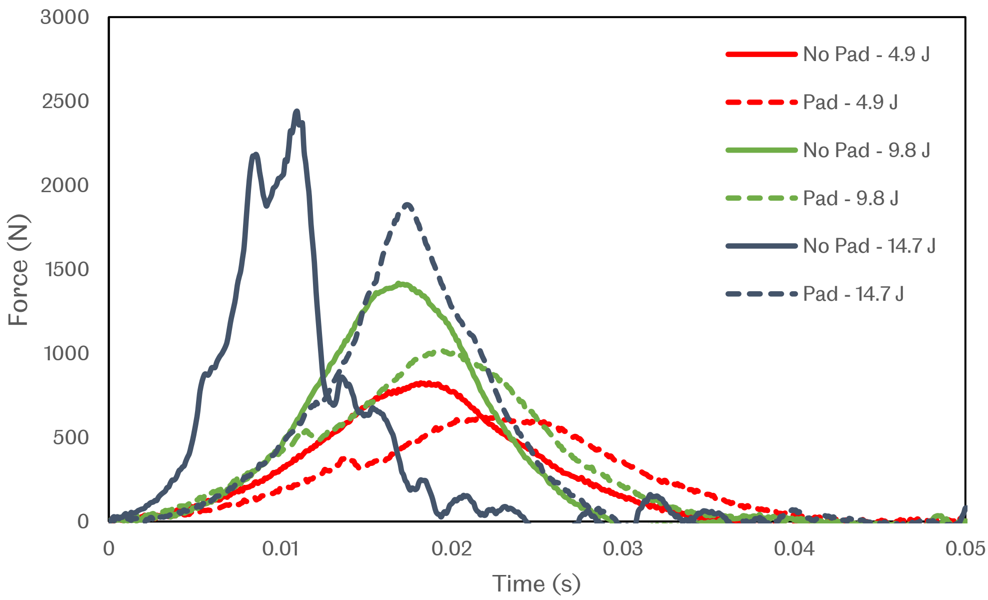


Figure 6.19 – Force-time trace at differing impact energies with and without padding impacting the Trapezius insertion into the shoulder.

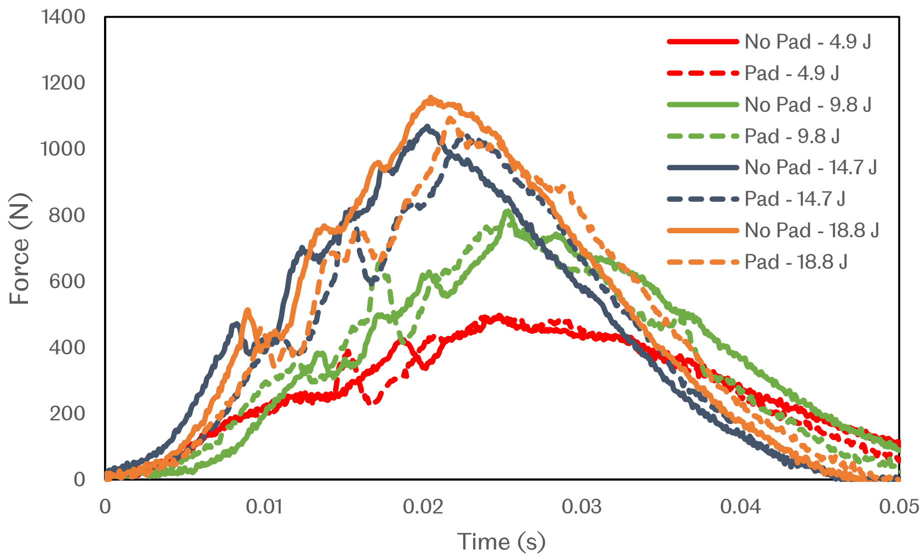


Figure 6.20 – Force time trace at differing impact energies with and without padding impacting the midpoint between the Acromion and the Seventh Cervical Vertebra on the Trapezius.

Addition of Commercial Shoulder Padding

It was also important to develop an understanding of how the impact response might vary when commercial shoulder padding was added to the shoulder surrogate instead of the control material Plastazote. The same procedures were followed as described above, however, two padding samples (samples 3 & 6) were impacted with the addition of the anatomical shoulder surrogate. The surrogate was impacted at the AC joint and middle belly of the trapezius illustrated in Figure 6.15.

Table 6.13 displays the peak force values (N) at differing energies as well as the mean force attenuation the padding is exhibiting calculated from the results displayed in Table 6.12. Similar force attenuation values are seen to that of Plastazote (Table 6.13). However, when impacting the AC joint at 9.8 J, only 12 % of the force is attenuated, compared with Plastazote which attenuates 45 %. This may be due to the padding samples bottoming out at this impact energy.

Table 6.13 – Peak impact force and force attenuation of commercial padding sample at two impact locations on the anatomical shoulder surrogate (Mean ± SD).

Impact Location	Energy (J)	2.3	4.9	9.8	14.7
AC Joint	Peak Force (N) – Sample 3	500 ± 25	2259 ± 128	7405 ± 343	-
	Peak Force (N) – Sample 6	460 ± 113	1878 ± 153	6973 ± 142	-
	Mean Force Attenuation (%)	73	62	12	-
Middle Belly of Trapezius	Peak Force (N) – Sample 3	-	363 ± 3	543 ± 8	725 ± 29
	Peak Force (N) – Sample 6	-	372 ± 4	559 ± 12	717 ± 27
	Force Attenuation (%)	-	26	32	32

When analysing Figure 6.21b, the gradient of the curve with the addition of MS 3 & 6 is far steeper than that of Plastazote. When analysing Figure 6.21a, the gradient of the curve is initially similar to that of Plastazote, however, Plastazote still attenuates more impact force in comparison.

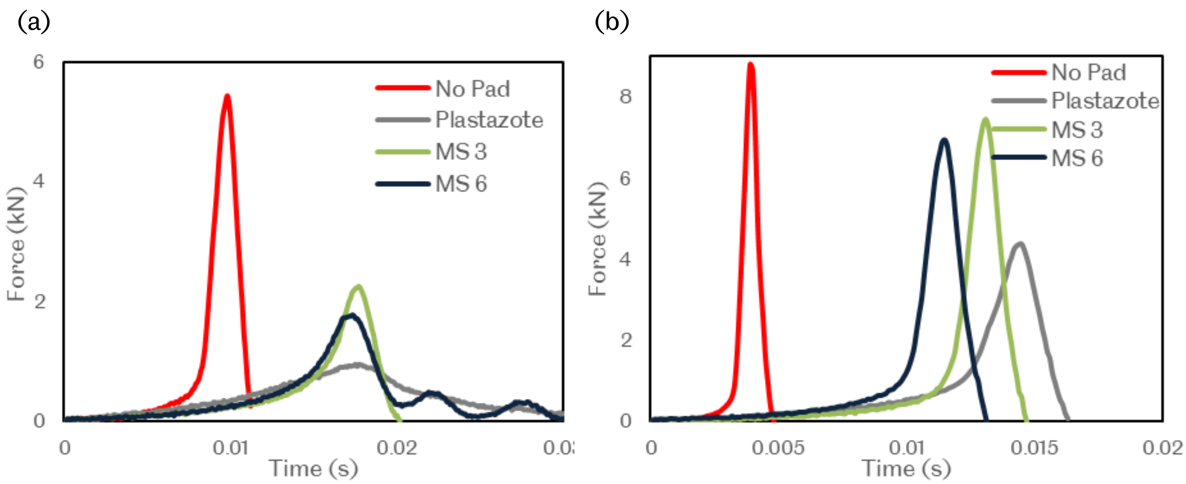


Figure 6.21 – Force time trace with and without commercial padding samples impacting the AC joint at (a) 4.9 J, (b) 9.8 J.

6.4.4 Discussion

It was outlined at the start of section 6.4 that its aims were to:

1. To determine current padded rugby clothing's ability to protect from Lacerations, Cuts, and Abrasions caused by bony parts of other players.
2. To determine the differences in impact response behaviour at different parts of the human shoulder.

This was split into two distinct studies, the study using the simplified shoulder surrogate where conclusions could be made regarding padded clothing's ability to protect from Lacerations, Cuts, and Abrasions caused by bony parts of other players and the study using the anatomical surrogate where conclusions could be made regarding the impact response behaviour at different parts of the human shoulder. Both these studies were completed so key findings could be made to satisfy World Rugby's™ RQ3.

When referring to the test set-up described in 6.4.2, it took an impact of around 7.7 kN to tear the surrogate with no padding, replicating a laceration injury. Forces in Rugby Union impacts range from 1 kN to 6.2 kN [61, 62, 221] so impacts of this magnitude tend to not be experienced. This may explain the low frequency (0 -21.24 injuries per 1000 PMH) of laceration injuries that occur in Rugby Union.

Regarding current commercial padding samples, it was found that they were attenuating on average 41 % of the force when impacted with the simplified surrogate and rigid dome striker. This was less when compared with the Regulation 12 set up (rigid anvil and rigid flat striker) of 62.7 % of force attenuated. The amount of force shoulder padding can attenuate in an impact in a rugby match may therefore be less than when tested in a laboratory, this should therefore be considered when devising new testing procedures for shoulder padding.

A 165 % increase in impact force was needed to cause a tear in the chamois leather skin of the simplified surrogate when shoulder padding was added, compared to without. This suggesting that the use of current shoulder padding will give around an additional 1.5 times the amount of protection to preventing Cuts and Lacerations from blunt force impacts in a game of rugby (Figure 6.14). When relating these results to RQ 3, this confirms shoulder padding does protect from Cuts and Lacerations in blunt impacts. More importantly, when recommending new test procedures to Regulation 12, this ability needs to be assessed. The best way to assess this would be by monitoring the force attenuation of the padding as assessed in §6.4.2.

Testing was completed using the anatomical surrogate and rigid dome striker. A key finding was the difference in force attenuation of shoulder padding at different locations on the shoulder. Commercial padding samples attenuated a far greater amount of impact force when impacted with the AC joint (62 %), a more severe impact due to the increased rigidity in the AC joint compared with the middle belly of the Trapezius (26 %) at 4.9 J. This suggests padding may offer little, to no protection in a less severe impact with much of the force transferring through onto the shoulder, this further backs up the findings of Harris & Spears [182] in a far more biofidelic impact scenario than they presented. However, caution should be taken as the shoulder surrogate represents a relaxed Trapezius muscle and when this is contracted, outcomes could differ. The peak force of the impact would likely be larger due to the increased rigidity in a contracted muscle compared with relaxed.

When assessing the impacts on the AC joint of the anatomical shoulder surrogate. The force attenuation of the commercial padding samples decreases as the impact energy is increased. This decreases dramatically from 62 to 12 % between impacts at 4.9 to 9.8 J. This is likely due to the foam in the padding 'bottoming out' as described in §6.3.4. If this occurs in a game of rugby, the loads transferred through into the shoulder could be high. This dramatic change in attenuation is concerning, the greater amount of attenuation provided at low loads may give a player an impression of invincibility or reduce the sensory feedback provided by the shoulder, however a tackle with a greater load may lead to shoulder padding offering a much-reduced amount of protection. What is reassuring is this did only occur at peak forces of over 8 kN, Which is 2 kN higher than the range in tackle forces (2 – 6 kN) found by Seminati et al. [62]. What needs to be considered when providing recommendations for updated test protocols is that shoulder padding can offer differing amounts of protection dependent on the shoulder impact location as well as the force.

6.5 Assessment of Stud Induced Impact Injuries

6.5.1 Introduction

Section 6.4 assessed padded clothing's ability to protect from Cuts, Lacerations, and Abrasions caused by other body parts on a rugby player. It was highlighted that these injuries could also be caused by rugby player's studs. It was therefore vital padded clothing's ability to protect from stud-induced injuries was assessed to fully answer World Rugby's™ RQ3.

As mentioned in §2.2 Oudshoorn [157] completed a doctoral thesis developing a test method for assessing the injury risk of studs during game-relevant loading conditions. It was found that stud impacts had two distinct phases. An initial direct impact and a raking phase. This section, therefore, looks to use the test parameters developed in Oudshoorn's research as a guide to developing test methods to assess the ability of padded clothing to protect from stud-induced injuries. The work completed in this section was split into two studies, In the first study, direct stud impacts were replicated using the drop rig and simplified surrogate described in §6.4.2, however, the striker was modified to a rugby stud. In the second study, a new test method was developed to replicate stud raking and is described in §6.5.3. These two studies encompassed the following aims and objectives:

1. To develop and validate repeatable test methods that will replicate relevant in-game stud loading conditions, to assess padded clothing's performance.
2. To determine current padded clothing's ability to protect from Lacerations, Cuts, and Abrasions caused by rugby player's studs.

6.5.2 Stud Impacts

Test Methodology

Impact testing using a stud tipped striker and simplified shoulder surrogate took place to simulate Cut/ Laceration injuries caused by stud impacts. An incremental process was used to find the impact energy and force the surrogate tore at, without padding and with Plastazote (control) and two padding samples (MS 1 & 2). The experimental test setup is shown in Figure 6.22.

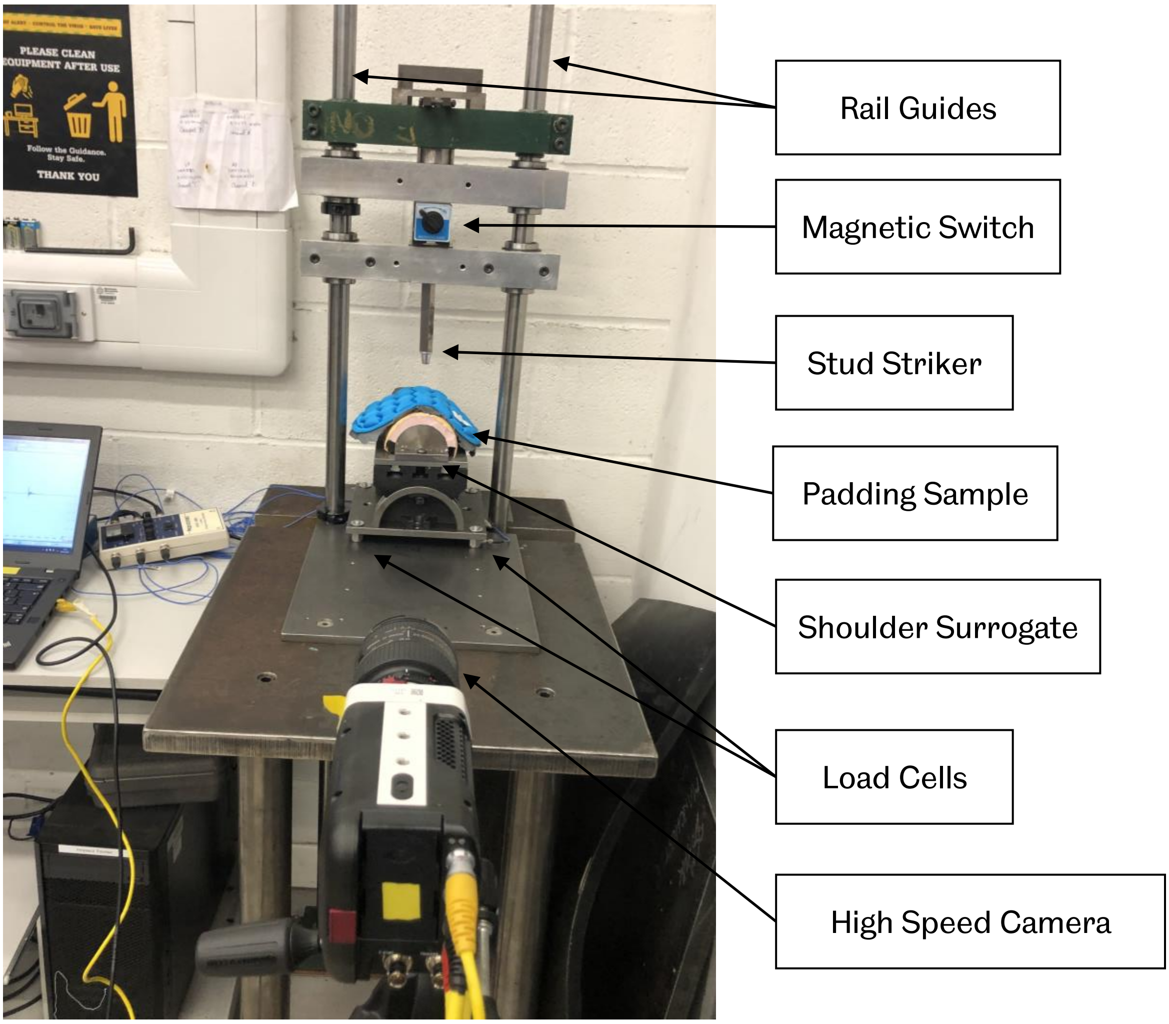


Figure 6.22 – Experimental test set up for replication of direct stud impacts using simplified shoulder surrogate.

The incremental process used was so that the minimum impact energy and force to cause a tear in the surrogate could be found. A tear was defined when the silicone soft tissue layer could be seen through the chamois skin.

The test procedures are listed below:

1. The stud striker (diameter = 11 mm, mass = 3.65 kg, World Rugby™ approved) was set to the desired drop height.
2. The striker was released by switching off the magnet and it impacted with the simplified surrogate.
3. The surrogate was removed from the drop rig and a picture taken, it was then put back into the drop rig, however, a different impacting location was chosen so no degradation from previous impacts was present.

4. The striker's drop height was incrementally increased and this process repeated until a tear occurred.
5. The same process was repeated but with the addition of Plastazote (control) and two padding samples over the surrogate.
6. Both a force-time trace and high-speed video were recorded to aid the evaluation of results.

Results

The surrogate was impacted in various locations with and without padding, peak impact force was attained, and the surrogate skin was examined for a tear (Y/N). High-speed video stills demonstrating the impacts conducted are shown in Figure 6.23.

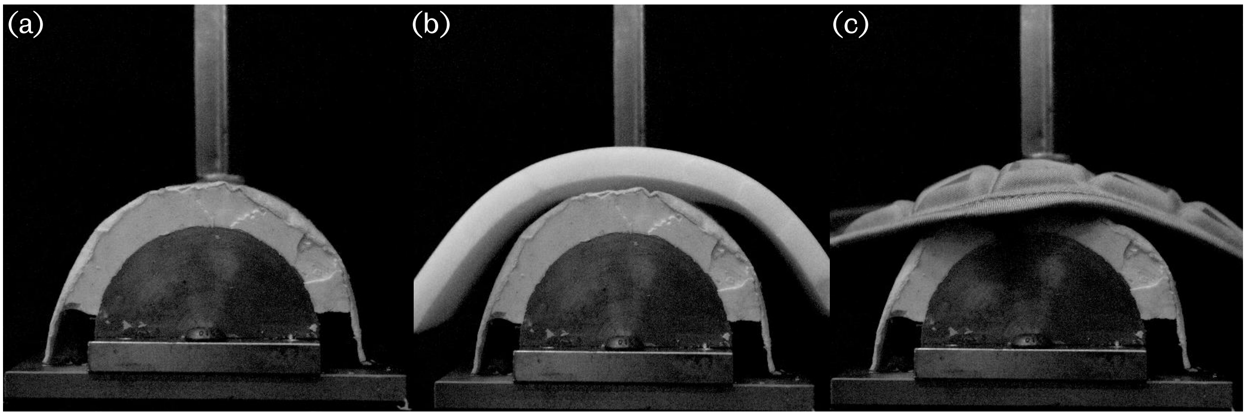


Figure 6.2 – High-speed video stills at max displacement (a) without padding, (b) Plastazote, (c) manufacturer padding sample.

The impact energy and peak force of each impact, as well as whether this impact caused a tear, have been reported in Table 6.14. Further impacts at energies that caused a tear with the addition of padding were also performed without padding. The images in figure 6.23 were taken at max displacement, this was measured using the High-speed video and calculated using equation 6.3.

Table 6.14 – Peak impact force and tear (Y/N) of simplified surrogate impact testing, with and without padding.

	Energy (j)	1.8	2.1	2.5	2.9	3.2	3.6	3.9	4.3	4.7	5.0
No Padding	Force (N)	545	731	1091	-	-	-	-	-	-	2432
	Tear (Y/N)	N	N	Y	-	-	-	-	-	-	Y
Plastazote	Force (N)	-	400	549	652	656	700	841	940	1154	1355
	Tear (Y/N)	-	N	N	N	N	N	N	N	N	Y
MS 1	Force (N)	-	-	195	-	214	1085	`1180	1585	1623	1724
	Tear (Y/N)	-	-	N	-	N	N	N	N	N	Y
MS 2	Force (N)	-	502	658	668	845	1070	1191	1259	-	-
	Tear (Y/N)	-	N	N	N	N	Y	Y	Y	-	-

A tear in the chamois skin is seen at 2.5 J with no padding, an increase of 1.1 J is needed to cause a tear with padding sample 2 added, and 2.5 J with Plastazote and padding sample 1 added. Images of the torn surrogate can be seen in Figure 6.24. A tear occurred when the silicone soft tissue layer could be seen through the skin. Analysis of the peak force values should be taken with caution as both the padding and the shoulder surrogate will be attenuating a certain amount of this force.

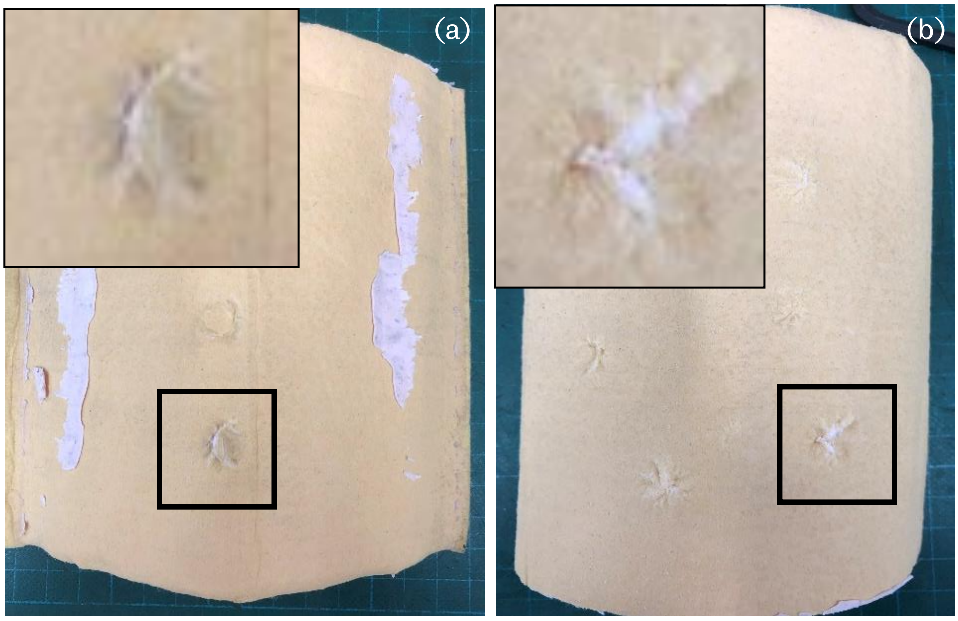


Figure 6.24 – Images of tearing caused by stud impacts (a) No padding, (b) Plus Plastazote.

The rationale behind the impact testing described was to gauge what level of protection to Cut and Laceration injuries caused by direct stud impacts padding samples are currently offering. Table 6.15 shows how much extra impact energy and force (as a percentage) is needed to tear the shoulder surrogate when padding was added when compared with the 'No padding' results.

Table 6.15 – Impact energy and force increase to exhibit tearing caused by studs with the addition of padding.

% Energy increase to tear with padding				% Force increase to tear with padding			
Plastazote	MS 1	MS 2	Mean	Plastazote	MS 1	MS 2	Mean
100	100	49	81 ± 33	24	58	-2	27 ± 30

A mean of an 81% increase in impact energy was needed to cause tearing. The percentage increase in force is lower (27 %), however, it should be noted that this would be because the padding is attenuating a certain amount of this force.

It was important to gain insights into the amount of impact force both the padding and the shoulder surrogate were attenuating. A similar process was used in §6.3.2 and §6.4.2 by where the peak impact force of rigid body impacts was measured at low energies and then extrapolated. This was then compared with FE model data developed by PhD A. Using this data, the amount of force attenuated by the surrogate and the padding could be measured. To do this, the soft tissue layers of the surrogate were removed, leaving the rigid bone layer. The stud impactor was then dropped on the rigid bone layer as seen in Figure 6.25.

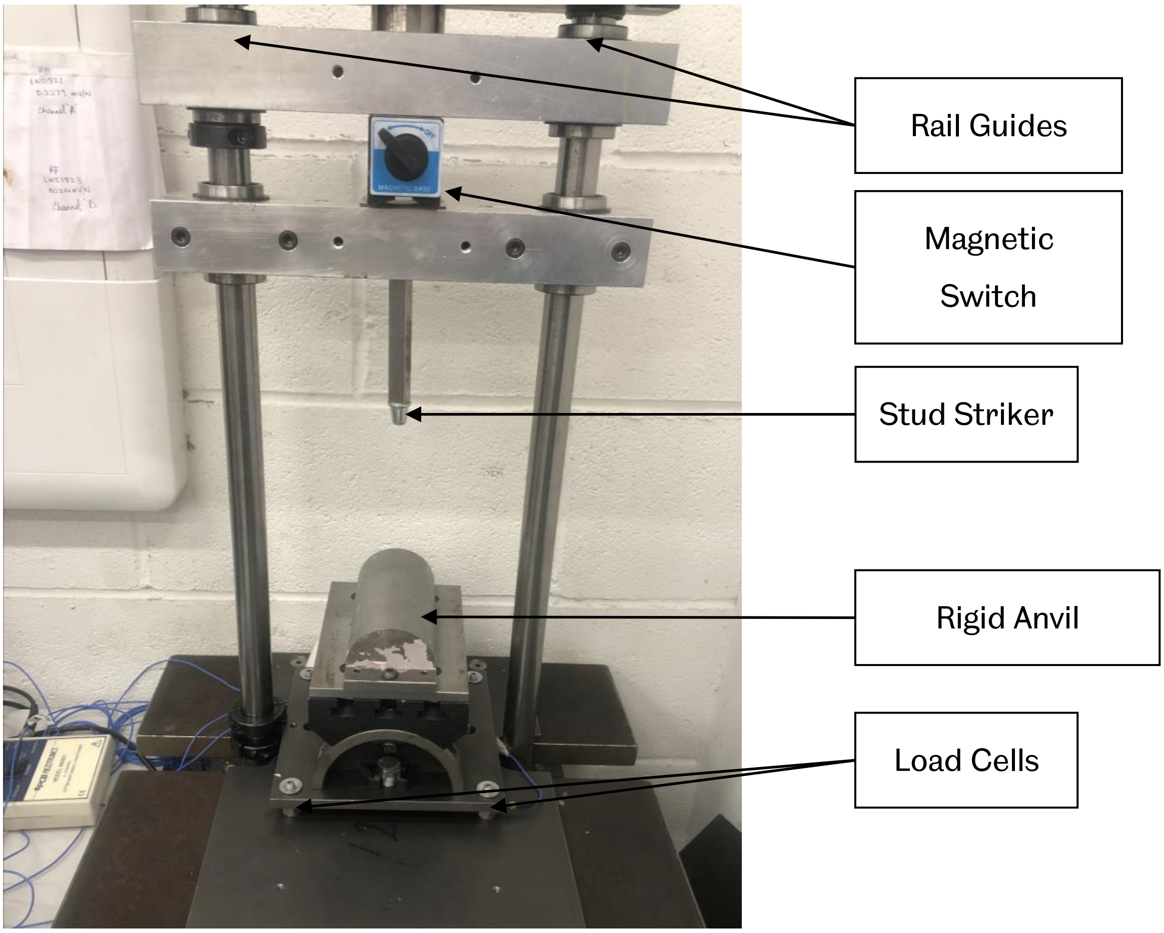


Figure 6.25 – Experimental setup for rigid body impact using stud striker.

Peak values can be seen in Table 6.16, a linear trend line was calculated (Figure 6.26) and peak force (N) at higher impact energies including when surrogate tearing occurred was estimated (Table 6.17). An FE model was also developed by PhD A. Peak force values and the percentage difference from extrapolated values are displayed. The average percentage difference was 0.4 %. Extrapolated peak force values were a far closer match to the FE model values than in §6.4.2. As with §6.3.2, the extrapolated values from the measured impacts at lower energies were used for later analysis.

Table 6.16 – Experimental rigid stud impact peak force values.

Impact Energy (J)	1.1			1.4			1.8			2.1		
Trial	1	2	3	1	2	3	1	2	3	1	2	3
Peak Force (N)	4140	4719	4850	5771	5600	5901	6853	6955	7234	8812	8616	9035
Mean	4569			5757			7014			8821		
SD	378			151			197			210		

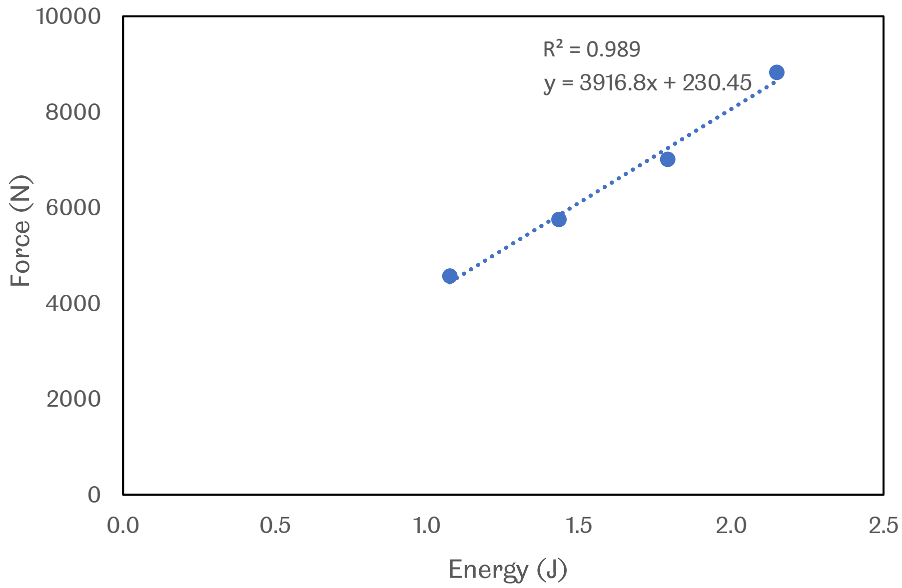


Figure 6.26 – Experimental peak force values with liner trend line.

Table 6.17 – Estimated rigid stud impact peak force values.

Impact Energy (J)	2.5	2.9	3.2	3.6	3.9	4.3	4.7	5.0
Extrapolated Peak Force (N)	10022	11589	12764	14331	15506	17073	18639	19814
FE Model Peak Force (N)	10026	11458	12890	14323	15755	17187	18619	20052
% Difference	0.04	1.13	0.99	0.06	1.61	0.67	0.11	1.2

Using the results outlined above, the amount of force the simplified shoulder surrogate and the padding is attenuating can be calculated. Once these calculations have been completed, further assessments on padded clothing's ability to protect from Cuts and Lacerations caused by studs can be made. The peak force of the rigid body impact at certain impact energies has been defined as the **dynamic impact force.** Table 6.18 displays the peak force values exhibited at 2.5 J (when tearing of the chamois skin occurred without padding) and at 5 J (when tearing of the chamois skin occurred with padding added). The percentage reduction in force has also been calculated so that the amount of force the surrogate and the padding are attenuating can be defined. The average of the forces between Plastazote, MS 1, and MS 2 was used.

Table 6.18 – Peak force and force attenuation at tearing with and without padding.

Impact Energy	Impact Test Set-up	Peak Force (N)	Reduction in Force from Rigid-Body Impact (%)
2.5 J (No Padding)	Rigid Body Impact*	10022	-
	Addition of the Surrogate	549	95
5 J (Padding)	Rigid Body Impact*	19814	-
	Addition of the Surrogate	2432	88
	Addition of the Surrogate and Padding	1539	92

*Peak force has been extrapolated from experimental data.

The results in Table 6.18 show 3 key results that can be used to assess the effectiveness of padding at preventing Cuts and Lacerations.

1. The surrogate is on average attenuating **91.5%** *((95 + 88) /2)* of the force compared to the rigid anvil.
2. Padding is on average attenuating **37%** *((2432 – 1539) /2432) * 100)* of the impact force with the addition of the surrogate and anvil.
3. An increase of **98%** (9792 N) *((19814 – 10022) / 10022 * 100)* in the dynamic impact force is needed to cause a cut/ tear when padding is added to the surrogate compared to when it is not. If we were to work backwards this would be a **49%** *((73560 – 28760) / 73560) * 100)* decrease.

From the results displayed in Table 6.18, there is a range in dynamic impact force where padding is protecting the wearer from a Cut or Laceration that would have been caused by a stud without the addition of padding, this is displayed in Figure 6.27 and highlighted orange.

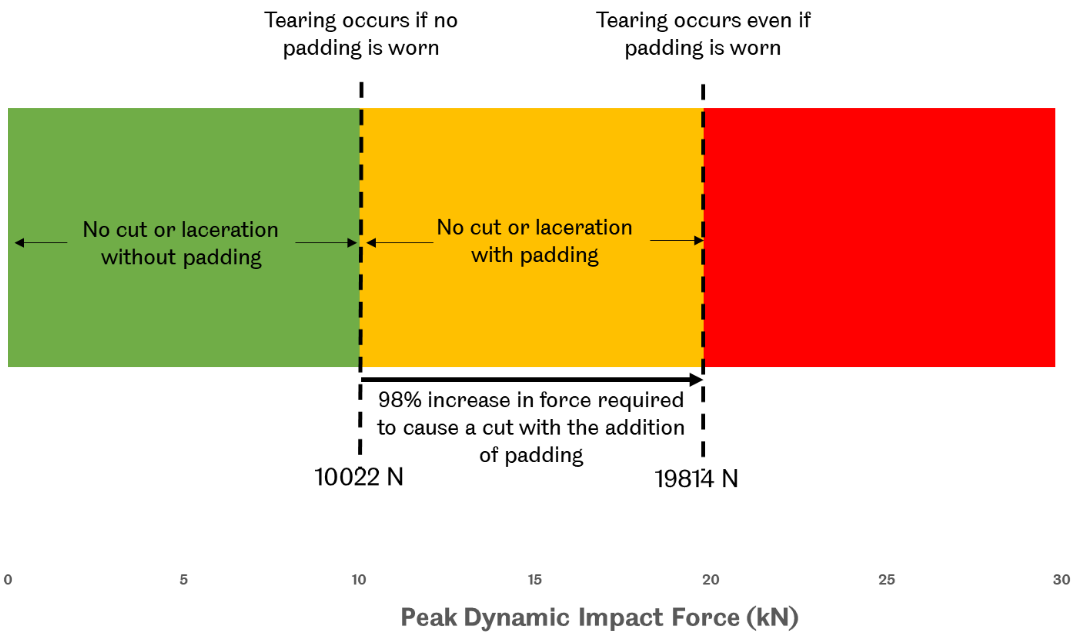


Figure 6.27 – Stacked bar chart of test results.

Repeatability of Testing

To ensure the results of the stud impact testing completed using the simplified surrogate were repeatable. The same testing procedures were completed one week after the original testing. The testing completed was to ensure that:

- Similar force-time traces including peak forces were exhibited.
- The surrogate tore at similar impact energies to the previous testing.

The same testing procedures outlined above were followed; however, fewer steps were taken in the incremental process to tearing. The results are outlined in Table 6.19. A tear in the surrogate was seen at the same impact energy as the previous testing for all the test conditions (No Padding, Plastazote, MS1, and MS 2). Similar impact forces were also recorded with % differences ranging from 0.5 – 28.5 %

Table 6.19 – Repeatability testing results.

	Energy (J)	1.8	2.1	2.5	2.9	3.2	3.6	4.3	4.7	5.0
No Padding	Force (N)	670	703	849	-	-	-	-	-	-
	% Difference	18.7	4.0	28.5	-	-	-	-	-	-
	Tear (Y/N)	N	N	Y	-	-	-	-	-	-
Plastazote	Force (N)	-	-	-	-	-	-	901	1098	1259
	% Difference	-	-	-	-	-	-	4.3	4.9	7.1
	Tear (Y/N)	-	-	-	-	-	-	N	N	Y
MS 1	Force (N)	-	-	-	-	-	-	1503	1573	1678
	% Difference	-	-	-	-	-	-	5.5	3.2	2.7
	Tear (Y/N)	-	-	-	-	-	-	N	N	Y
MS 2	Force (N)	-	-	-	689	841	1054	-	-	-
	% Difference	-	-	-	3.0	0.5	1.5	-	-	-
	Tear (Y/N)	-	-	-	N	N	Y	-	-	-

* % Difference is calculated from the results displayed in Table 6.14.

These results provide good support that the testing completed with the simplified surrogate and stud striker is repeatable. This means the results from this can be used to assess the performance of padded rugby clothing in relation to its ability to prevent from stud induced injury. This is further discussed in §6.6.

6.5.3 Stud Raking

6.5.3.1 Overview

This section has been adapted from a published submission to the Sports Engineering Journal.

Hughes, A. C., Dixon, J., Driscoll, H. F., Booth, J., & Carré, M. J. (2022). Padded rugby clothing to prevent laceration and abrasion injuries from stud raking: a method of assessment. Sports Engineering, 25(1), 1-8.

The roles of the other authors for this paper in relation to the project are as follows: M. Carré (supervisor: academic), H. Driscoll (supervisor, academic), J. Dixon (Masters Student). The Manuscript was written by A. Hughes, all authors commented on the manuscript. 90% of the research and 100% of the writing was done by A. Hughes.

Abstract

Padded clothing (shoulder padding) is worn in Rugby Union to give players an opportunity to protect themselves. A performance specification for padded clothing has been set out by World Rugby™, with the intention that padded clothing only protects against cuts and abrasion. Test protocols in this specification provide an assessment of the impact force attenuative properties of the material, this itself will not indicate what injuries they may have the potential to prevent or lessen the severity of. The current study has used previously established biomechanical parameters to develop a mechanical test procedure to assess the ability of padded clothing to prevent or lessen the severity of stud-induced injuries. A synthetic skin and soft tissue surrogate was developed and validated to mimic human anatomy. Without the addition of padded clothing, both wearing (abrasion) and tearing (laceration) of the synthetic tissue surrogate were seen. The addition of padded clothing saw no sign of stud-induced injury, even after six repeated trials of the same product, showing padded clothing has the ability to prevent or lessen the severity of superficial injuries such as lacerations and abrasions. The developed testing protocols have the ability to assess both the safety of any sports studs as well as the effectiveness of various protective clothing products across the sports industry.

6.5.3.2 Aims and Objectives

The purpose of this research was to develop a new test procedure using appropriate loading conditions to assess the effectiveness of rugby shoulder padding to reduce the severity of Lacerations and Abrasions induced by stud raking contacts.

6.5.3.3 Design of Rig

To simulate rugby stud raking a rig initially developed to assess shoe-surface interactions [222] was adapted to include a rugby stud attachment as pictured in Figure 6.28. This rig allowed for the replication of rugby stud raking conditions previously investigated by Oudshoorn [158]. Both vertical and horizontal force can be applied using pneumatic cylinders, the pressure can be altered to modify loads to suit. The rig is instrumented with load cells and linear variable differential transformers so that force, displacement, and acceleration in both the vertical and horizontal directions can be measured. The rig had a minimum vertical force (~ 500 N), therefore four studs (Gilbert, 18 mm, aluminium (Conforming to regulation 12, schedule 2 [223])) were used in a configuration that matched the front four studs of rugby boot (Adidas, Kakari, SG), so that similar pressures to Oudshoorn's [158] parameters (one stud) could be achieved.

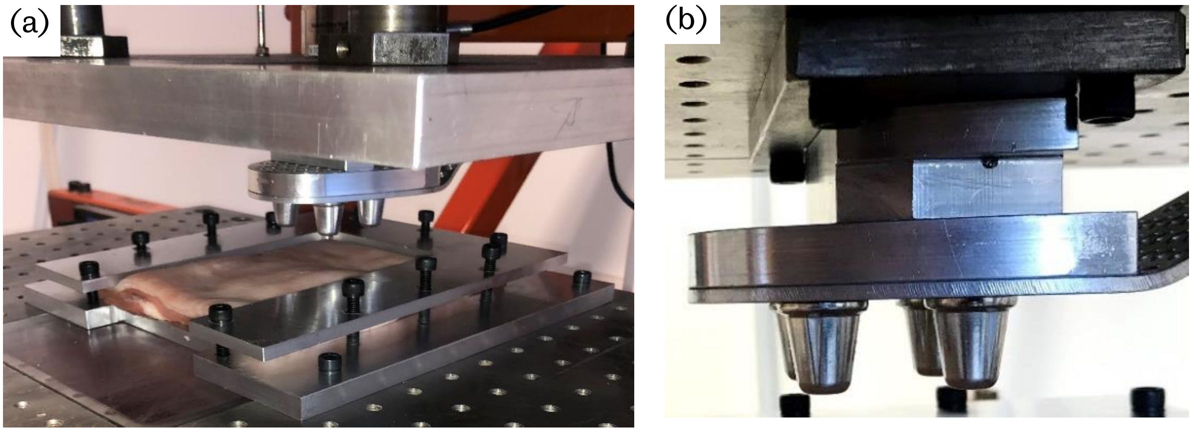


Figure 6.28 - (a) Test rig with porcine tissue testbed attached; (b) Rugby stud (18mm) attachment.

A synthetic tissue surrogate was developed to replicate the human soft tissue response to stud raking (Figure 6.29). It consisted of two layers, simulating muscle and skin. The dermal and epidermal layer of skin combined is 1.93 – 2.35 mm thick [224]. The trapezius, which is where shoulder padding sits on the rugby player ranges from 4.1 – 14.26 mm in thickness [225, 226]. A 2 mm skin layer and a 8 mm muscle layer (10 mm total) were therefore used for the synthetic tissue surrogate. Both the validation simulants (Porcine tissue and Syndaver® skin simulant) were also 10 mm thick. The muscle simulant was made from silicone (Silastic™ 3481, Dow Corning, UK) set as a rectangular slab (200 × 150 × 8 mm). The reason a flat rectangular shape and not a shoulder shape was chosen was so the surrogate was repeatable, easy to fabricate, and fitted into the test rig. The silicone was a three-part blend with the addition of a catalyst and deadener (PlatSil® Gel 25 Deadener, Mouldlife, UK) in a 10:1:4 (base: catalyst: deadener) weight ratio. The blend was thoroughly mixed and fully degassed before being poured into rectangular moulds (200 x 150 mm). Its compressive properties matched that of Porcine muscle tissue, with previous research [227] outlining how this was achieved. The skin layer was made from synthetic chamois (2 mm) cross-woven polyvinyl acetate (PVA) (KCIC200, Kent Car Care, Manchester, UK) due to its similar penetration resistance to the skin as found in previous studies [124, 125, 158]. The skin covered the soft tissue layer to create a dual-layered testbed (10 mm thick in total). The chamois was added to the silicone while it was setting so they bonded together. A bespoke clamp held the skin-tissue system in place during the testing (Figure 6.29). Padded rugby clothing (shoulder padding) can also be clamped in place over the synthetic tissue surrogate. The test rig operates in a two stage process. A vertical load applies pre-compression to the surrogate following which a horizontal load is applied to create a raking action with the vertical load maintained.

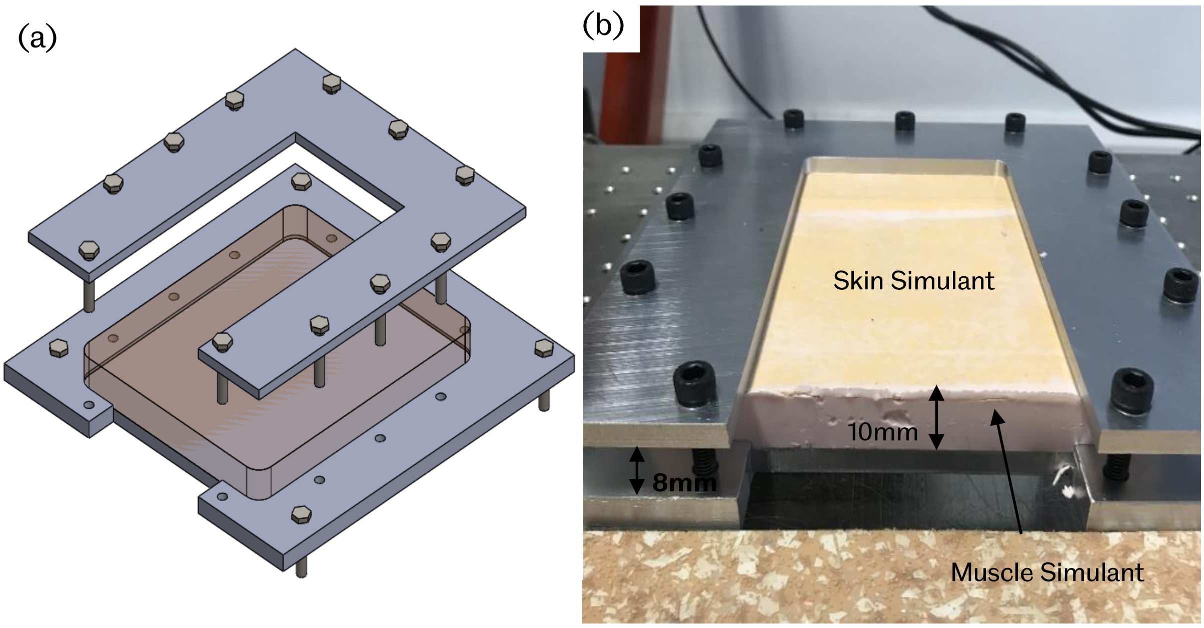


Figure 6.29 - (a) – Exploded CAD view of Test Bed, (b) – Test Bed (without padding).

6.5.3.4 Validation of Protocol

Test Parameters

Using the adaptations described to the rig and the surrogate, similar loading conditions developed by Oudshoorn [157] were achieved (Table 6.20). It should be acknowledged that maximum force (N) and therefore pressure per stud (MPa) is higher in the current test. However, this is still within 95 % CI of Oudshoorn's study.

Table 6.20 - Test parameters (without padding) ± Standard Deviation (SD).

	Max. Vertical Force (N)	Max Pressure (MPa) (per stud)	Mean Horizontal (Raking) Velocity (m/s)	Horizontal Force at Max. Vertical Force (N)	Mean COF
Oudshoorn [158] (1 stud)	137 ± 39	1.32 ± 0.38	0.93	-	0.57
Current Test (4 studs)	729 ± 1.1	1.75 ± 0.003	0.82 ± 0.06	641 ± 0.8	0.63

Synthetic Tissue Surrogate

To validate the synthetic tissue surrogate, tests were run on, i) a slab of 10 mm thick ex vivo Porcine tissue (belly) acclimatised to room temperature (stored in a fridge (4 °C)) and ii) a slab of commercially available synthetic skin and subcutaneous tissue ("Basic" 5 mm thick tissue plate, Syndaver®, Florida) covering a 5 mm thick silicone muscle layer (same as muscle layer in synthetic tissue surrogate). The procurement and storage of the Porcine tissue followed all of the University of Sheffield's ethical guidelines and risk assessment policies. This approach offered a multi-factor validation of the surrogate. Porcine tissue and Syndaver® products have been reported to display similar mechanical properties to human tissue, especially when comparing penetration resistance of the skin [228]. While Syndaver® skin is considered to be a state-of-the-art product, chamois leather offers a low-cost and frangible alternative that is suitable for this project.

Raking tests (without the addition of shoulder padding) following the parameters in Table 6.20 were performed on the Synthetic tissue surrogate. And then the Porcine tissue and Syndaver® skin simulant for validation. When comparing post-test photographic images (taken from a fixed camera position) (Figure 6.30), similar signs of wear/ abrasion were seen between the synthetic tissue surrogate and Porcine tissue, as well as this, there was tearing/laceration in the Syndaver® product, as seen in the synthetic tissue surrogate. A tear/laceration was defined when the silicone soft tissue layer could be seen through the chamois layer. The reason the Porcine skin did not tear could be because it is slightly tougher due to its higher collagen content [118, 229], and slightly higher thickness when compared with human skin [230]. When comparing vertical load against horizontal displacement plots (Figure 6.31) similar traces can be seen between the synthetic tissue surrogate and the Syndaver® skin simulant, the Porcine tissue surrogate follows a similar trace at the end of the raking movement but the loads are smaller at the beginning, possibly due to the irregularities seen in organic tissues. After validation, the synthetic tissue surrogate was used for the assessment of padded clothing.

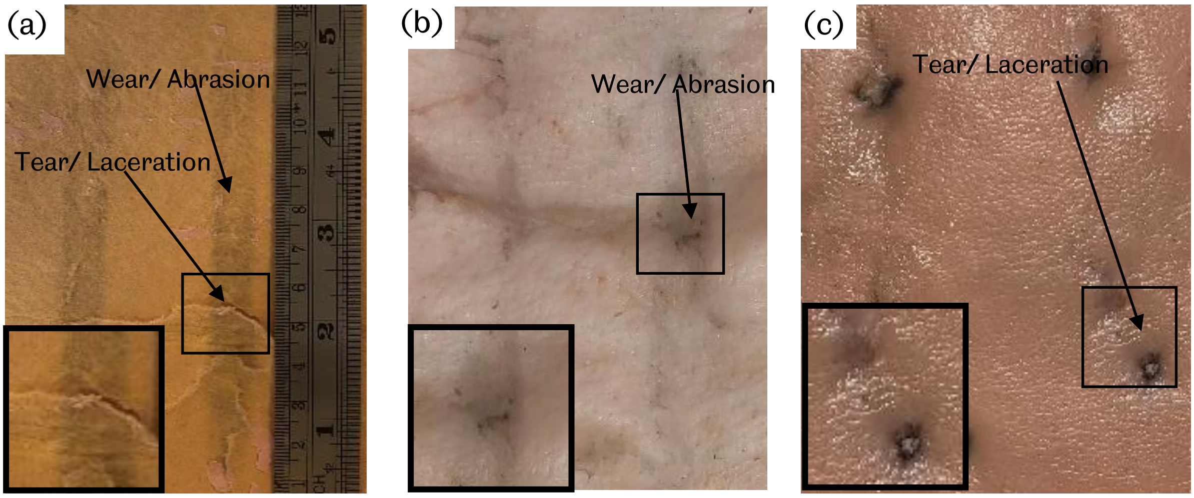


Figure 6.30 - (a) - Synthetic tissue surrogate, (b) - Porcine tissue, (c) - Syndaver® skin simulant.

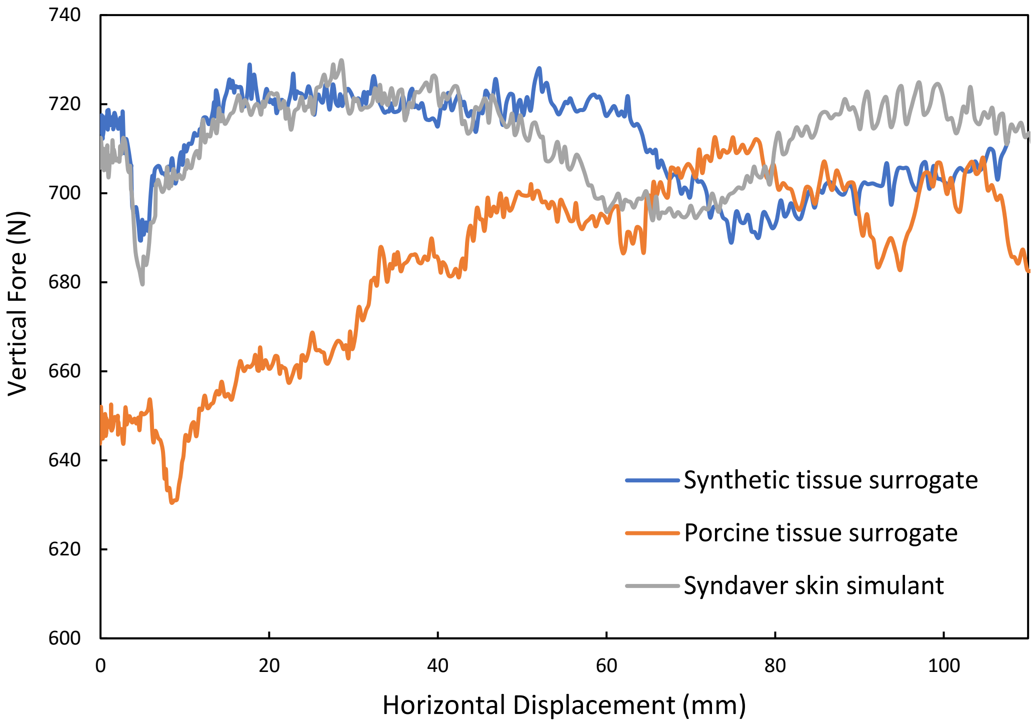


Figure 6.31 - Vertical load (N) against horizontal displacement (mm) plot for each validation surrogate.

6.5.3.5 Assessment of Padded Clothing

Testing was completed with the addition of PlastaZote foam (PlastaZote LD60, 12 mm thick) used as a control material, as done in a previous impact testing of shoulder padding study [231], and two commercial shoulder padding materials clamped over the synthetic tissue surrogate. Images of the synthetic tissue surrogate were taken before and after one raking trial. After this, a further five (six in total) raking trials were performed to see if the padding degraded. There was clear degradation of both the padding, and the material that encompasses the padding, but no wear or tear marks on the underlying synthetic tissue surrogate (Figure 6.32a). The maximum vertical force applied without padding was 21 N (3 %) higher than the overall mean with padding, the mean horizontal velocity was 0.11 m/s (12 %) lower than the mean with padding, however, these differences were not significant (t-test, $p = .137$) (Table 6.21).

Table 6.21 - Testing Parameters and Outcome (Laceration).

	Max. Vertical Force (N)	Average velocity (m/s)	Laceration (Y/N)*
No Padding	729 ± 1.1	0.82 ± 0.06	Y
PlastaZote Foam	695	0.95	N
Commercial Pad 1	706	0.87	N
Commercial Pad 2	723	0.97	N

*Y = yes, N = no

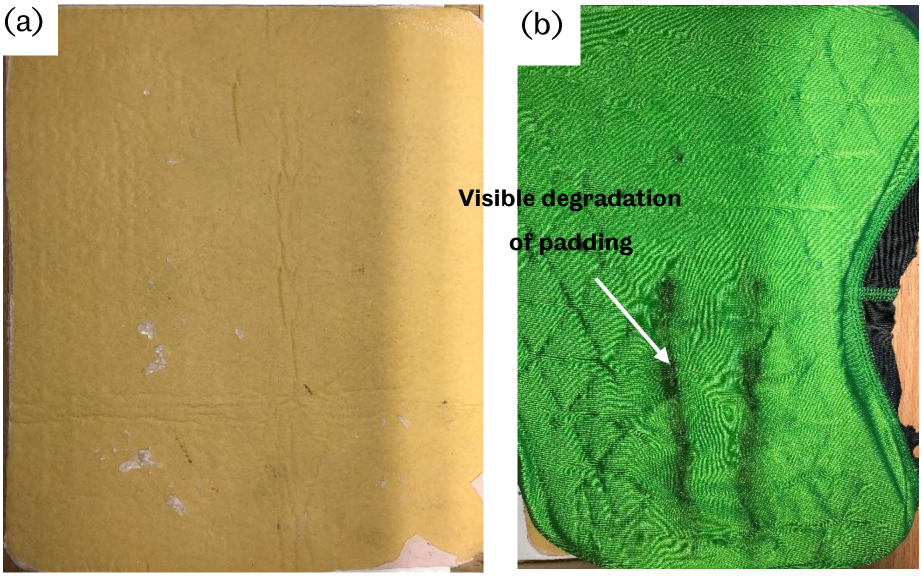


Figure 6.32 - (a) Synthetic tissue surrogate after 6 raking tests (b) Commercial padding 5 after 6 raking tests.

Figure 6.33 displays a difference map created by comparing images of the synthetic tissue surrogate before and after raking (ImageDiff, Ionforge). The software uses a colour scale to highlight differences between the pixels in images (purple < blue < green < orange < yellow). The wear mark seen in Figure 6.33a totalled 1108 mm^2 (left) and 988 mm^2 (right). The green areas are where a visible tear occurred (Figure 6.30a), as can be seen on the original image taken after raking (Figure 6.33c). There was minimal difference between the images taken before and after testing with the control material (Figure 6.33b), providing further evidence to suggest padding can prevent, or reduce the severity of, stud-induced abrasions or lacerations.

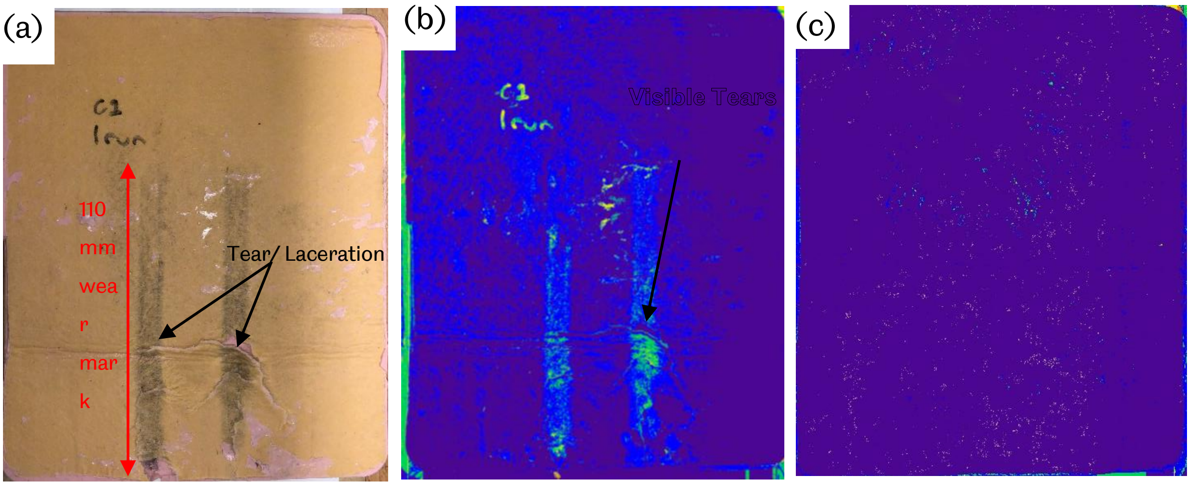


Figure 6.33 - Difference map created from an overlay of surrogate images before and after testing with (a) Original image of synthetic tissue surrogate after raking test with no padding (b) no padding (c) Control material.

6.5.3.6 Discussion

The study set out to develop a testing mechanism using appropriate loading conditions to assess the effectiveness of rugby shoulder padding's ability to prevent or lessen the severity of lacerations and abrasions. The investigation showed that rugby shoulder padding has the ability to prevent stud lacerations and abrasions using representative stud raking parameters. The study also developed a test method that can be used to assess both the safety of any sport's studs as well as the effectiveness of various protective clothing across the sports industry.

Some limitations should be acknowledged when referring to both the test methods and results. Issues arise when using skin and human tissue simulants in an attempt to replicate the mechanical behaviour (frictional properties, breaking loads) of the human tissue it is representing. The current test method validates the developed skin and human tissue simulant using Porcine tissue, a state-of-the-art skin simulant (Syndaver®), and past literature. Using this validation process the most replicable and affordable simulant was developed, this adding to Oudshoorn's [158] previously developed test method. However, unless live human tissue is used, the exact mechanical response to stud impacts cannot be known. Porcine tissue is easy to obtain and is used because of its similarity to human tissue [232]. However, biological samples are unhygienic if not stored and transported correctly, quickly degrade, and are highly variable from sample to sample. Their inconsistency in mechanical properties is shown in Figure 6.31. The surrogate developed was flat to ensure repeatability in testing, although layer thicknesses were similar to that of a human shoulder, the surrogate's anatomical geometries were not, therefore the surrogate was not shaped like a shoulder. This may have affected the severity of the tears at different points in the surrogate. However, it is suggested the same conclusions on the effectiveness of padding to prevent laceration and abrasion injuries would be found. The purpose of the surrogate was to test shoulder padding's ability to prevent cut and abrasion injuries in a repeatable way, which it successfully does, however, the addition of a surrogate with more representative geometries could further validate the flat synthetic tissue surrogate. Furthermore, the muscle thickness parameters were taken from a general population. Rugby players will tend to have a larger muscle mass than the general population [181].

Oudshoorn's [158] impact parameters were based on a data set of participants with a mean mass of 76.2 kg. This is far lighter than the average professional rugby player (99.2 kg) [59]. Stud raking forces in professional rugby may then be greater, however, slightly higher stud pressures were used for the current test which may act to balance this out. Future studies should increase the stud pressures to a point where an abrasion or laceration is caused when shoulder padding is added. The protective limits of current shoulder padding could therefore be established.

Quantitative assessment of the amount of damage to the synthetic tissue surrogate also poses a limitation. Many similar studies only qualitatively assess the difference in images, i.e. is there a tear or not. The current study looks to bridge this gap through the use of an image difference software. However, this could be improved, 3-dimensionally scanning the surface of the synthetic tissue surrogate before and after stud raking would allow for the size of a tear to be quantitatively assessed, a similar method was used by Kalbermatten et al. [233] where a laser scanner was used to measure the percentage change in skin surface area of facial lacerations. If future tests were completed and tearing occurred, classifying each tear following a skin tear classification system like the Skin Tear Audit Research [234] would lead to an improved assessment of the results. This could be the case if the padded material was of worse quality, or the stud used for testing was sharper. The results from the current study do however mean that each tear did not need to be classified as the addition of padding did not cause a tear at all.

Mechanical test setups like this also allow for the comparison of results measured from different shoulder pads, therefore meaning the test could be implemented in test houses as part of regulations and standards [235]. World Rugby's™ current test method assesses the impact attenuative abilities of the material and not its injury prevention capacity. Coupled with this, the test has not been based on any biomechanical parameters. However, when designing mechanical tests in regulations and standards it is important to consider their practicality for test houses. The current study would not be appropriate for use in a test house due to the number of replicable skin and soft tissue simulants that would need to be used and the bespoke nature of the mechanical testing device. Further work could consider whether standardised test methods for cut-resistant materials (e.g. ISO 13997 and BS 388) can be applied to rugby padding, by comparing outputs from such standards with those from the tests presented here.

Future studies are needed to identify how different shoulder padding designs, different materials as well as the addition of the player's jersey can affect their ability to protect from stud injury. This information could then be used by manufacturers, as well as in the development of regulations associated with shoulder padding.

6.5.3.7 Conclusions

In this research, a test method was developed, which assessed rugby shoulder padding's ability to prevent or lessen the severity of rugby stud injuries. Previous research was used to reproduce game-relevant loading conditions for stud raking in Rugby Union. A validated skin and soft tissue simulant was used, advancing previous research. The test method also can be modified to assess various items of padded clothing across multiples sports. Future research can use this test method to quantify the protective abilities in regards to stud injuries of various shoulder padding designs.

6.6 Discussion

6.6.1 Performance of Padded Rugby Clothing

The research completed in this chapter provides good quality data for an understanding of the performance of padded rugby clothing, specifically its ability to prevent Cut, Laceration, and Abrasion injuries. Furthermore, this data can be used to help answer RQ3.

The impact force attenuation of protection in sport is a good predictor of its injury preventative capabilities [104, 219, 220]. A key finding in this work is that shoulder padding force attenuation characteristics can change dramatically depending on the severity of the impact. With a more severe impact exhibiting an increase in the percentage force attenuation. When impacting padding samples at 9.8 J using the current Regulation 12 set up (rigid striker and rigid anvil) the mean force attenuation was 60 %, but when the simplified shoulder surrogate was added to make a less rigid impact (rigid striker and shoulder surrogate anvil) the mean force attenuation dropped to 41 %. This is something that should be considered when developing new testing protocols. This becomes even more interesting when considering the complexity of the shoulder, as impacts in different locations could cause differences in the severity of the impact as discussed in §6.4.4 (impacts with rigid striker and anatomical shoulder surrogate). When converting this into a game situation, a player may have more protection from shoulder padding if the impact contacts the AC joint rather than the Trapezius. However, the force attenuation properties of muscle and the increased amount of soft tissue may also protect its underlying structures.

This section replicated three different injurious scenarios; Cuts and Lacerations by blunt force injury from another body part, Cuts, and Lacerations from direct impact with a stud and Cuts, Lacerations and Abrasions by raking contact with a stud. Current commercial padding samples did show an ability to prevent Cuts, Laceration, and Abrasions, which would have been obvious when placing a 10 mm thick piece of foam between the skin and impacting element. However, quantifying by how much is the key novel result that can be used to guide future regulations here. When assessing shoulder paddings' ability to prevent blunt force Cuts and Lacerations a 165 % increase in impact force was needed to cause a tear in the chamois leather skin of the simplified surrogate when shoulder padding was added. This can be compared with shoulder paddings' ability to prevent Cuts and Lacerations caused by a stud where a 98 % increase in impact force was needed to cause a tear in the chamois leather skin. Current commercial shoulder padding is therefore providing at least double times the amount of protection compared to not having any at all. When providing recommendations for new regulations, the ability to measure and quantify this in testing procedures is paramount. Although Cut and Laceration injuries are prevalent, consideration must be taken as they rarely occur on the shoulder and more on the lower extremities and the head. However, quantifying shoulder padding ability to prevent from Cut and Laceration injuries was paramount to answering RQ3. §6.5.3 assessed shoulder paddings' ability to prevent Cut and Abrasion injuries caused by the raking movement of a stud. It replicated typical loading parameters as performed by rugby players boots in a ruck. Cut and Abrasion injuries were seen without the addition of shoulder padding, when this was added no Cut or Abrasion injuries occurred. It is clear shoulder padding will prevent Cut and Abrasion injuries occurring at game-relevant raking loads, but by how much is still unknown; quantifying this would be the next step in the research.

The ability of padded clothing to prevent injuries such as Contusions and Fractures was not explored in this study, as they were not included in the original aims and research questions set out by World Rugby. However, the test protocols described in chapter 6 could be used to assess this. Impact force and impactor area could be converted to pressure and key parameters that cause Contusion injuries as used by Desmoulin & Anderson [102] for example, to quantify shoulder paddings' ability to prevent this injury.

6.6.2 Effectiveness of Impact Testing Setups

The impact testing configurations described in chapter 6 provided time and cost-efficient experimental setups that satisfied the requirements and provided data to answer the aims and research questions stated by World Rugby™. The impact drop rig could be modified easily to change the impacting striker as well as house different shoulder surrogates. The raking rig employed similar 'in game' raking parameters and the test protocols used could be utilised in other industries outside of Rugby Union.

More importantly, the testing protocols used to assess shoulder padding's performance were more biofidelic and representative of real-world loading conditions than previous regulations or any research that looks to quantify padding's protective capabilities. The use of shoulder surrogates made from human tissue simulants that could be integrated into an impact testing setup meant that similarities in the following as set out in §3.3.3 could be seen:

- The magnitude and rate of deformation of the body segment.
- The magnitude of stress and strain in the body segment.
- The extent of damage that may be caused to the body segment.
- The proportion of strain energy absorbed by the shoulder padding and body segment.
- The interaction of the shoulder padding with the body segment and subsequent distribution of pressure.
- The magnitudes of stress and strain experienced by the shoulder padding.

However, the testing setups did embody some simplifications and limitations. The dome-shaped striker used for blunt force impacts was representing bony body parts like other rugby players' shoulders, knees, or elbows. However, this was made from steel and therefore differences in the stiffness of the striking element would be seen. Adaption of the dome-shaped striker so it had a small soft tissue and skin layer over the steel rigid element could act as a way of making the test configuration more representative.

The shoulder surrogates in the drop rig setups were constrained to the anvil below them. Therefore, it was only the striking element that would move within the configuration. In a rugby impact, the shoulder could be moving in any direction, therefore changing the loading condition in the impact. A testing setup used by Payne [159] where a pendulum impact configuration was adopted, whereby a thigh surrogate was suspended from an A-frame using strings; could act as a more accurate scenario of an impact in rugby providing thigh surrogate was replaced with a shoulder surrogate.

The testing configurations could never be fully validated. To fully validate them, testing would need to be completed on live humans. It would however be unethical to cause intentional injury to a human for the purposes of this research. The next best alternative would be to use PMHS, however, issues regarding licensing for use as well as a lack of muscle tonicity do arise with this.

The testing setups were more biofidelic than the testing protocols described in the current regulation (§3.3). However, if these setups were used in a standard or regulation, issues such as cost, and repeatability would arise. This should be considered when recommending updated regulations to World Rugby™.

6.6.3 Key Data and Conclusions

Making clear exactly what data obtained in chapter 6 is essential in making recommendations for updated testing protocols to World Rugby™, and by doing this answering their research questions. This has been presented in Table 6.22.

Table 6.22 – Key data and conclusions and impact on recommendations to World Rugby™.

Section	Key Data and Conclusions	Influence on Recommendations
6.2.4	Impacts at lower energies (4.9 & 9.8 J) using the current Regulation 12 testing protocols cause impact	When recommending new testing procedures, the impact forces should represent that found in real-world

	forces more replicable of an impact in rugby.	scenarios. Lowering the impact energy to reflect these forces is crucial.
6.3.2	Current commercial shoulder padding is attenuating 41 % of peak impact force (**blunt force impacts**) and a 165 % increase in the dynamic impact force is needed to cause a cut/ tear when padding is added to the surrogate compared to when it is not.	Force attenuation can be used as a measure of padding protective capabilities and be used to set performance limits in updated testing protocols.
6.4.2	Current commercial shoulder padding is attenuating 37 % of peak impact force (**direct stud impacts**) and a 98 % in the dynamic impact force is needed to cause a cut/ tear when padding is added to the surrogate compared to when it is not.	Force attenuation can be used as a measure of padding protective capabilities and be used to set performance limits in updated testing protocols.
6	The development and fabrication of impact surrogates are time-consuming, costly, and not always repeatable.	Recommending the use of impact surrogates should be taken with caution. Testing houses want protocols that are repeatable and cost-effective. Using the data found in chapter 7 and transferring it to a test that is repeatable and easy to use in a test house, while also using it to set performance limits is crucial.

CHAPTER 7 - RECOMMENDATIONS FOR THE IMPROVED ASSESSMENT OF PADDED CLOTHING IN RUGBY UNION

7.1 Chapter Overview

This chapter presents the recommendations made to WR that were directly linked to the work completed in this project. The recommendations made related to the impact test protocols and performance requirements of padded rugby clothing. The chapter breaks each recommendation down, giving a rationale backed up by the engineering and science presented in the project. It should be noted that other recommendations were made, but were a direct result of the work completed by PhD A. These were therefore not included in this chapter. The final part of the chapter also makes recommendations that are not specific to the Regulation 12 project, but give suggestions relating to what other work could be completed from the testing protocols developed in this doctoral thesis.

7.2 Recommendation 1: Impact Attenuation Test

The current test protocols for the impact attenuation test are replicated in §3.3.2, this includes impact testing at a 14.7 J impact energy (5 kg flat impactor from 0.3 m) using a rigid impactor and anvil. The peak impact acceleration recorded must be > 150 g to pass. It is recommended that:

- The drop height is lowered to 0.2 m to reduce the impact energy to 9.8 J.
- The peak impact acceleration pass limit should be reduced from > 150 to > 100 g (4900 N).
- Peak force should be measured using a load cell rather than peak acceleration with an accelerometer.

The reasons for this are outlined below:

1. Reduction in impact energy to 9.8 J:
    a. When replicating Regulation 12 test protocols at 14.7 J on all padding samples an average peak force of 10689 N was seen (20°C). This is far greater than the highest tackle forces (5.3 ± 1.0 kN) found by Seminati [221]. When replicating Regulation 12 test protocols at 9.8 J on all padding samples an average peak force of 7745 N was seen (20°C). Although still greater, this is closer to the tackle forces found by Seminati. The pass limit of > 4900 N is also smaller than this.

    b. Although it is recommended the impact energy is lowered to 9.8 J, the test protocols should remain the same. A rigid anvil and striker should be kept as repeatability and accuracy of testing becomes an issue, especially in a test house, if a silicone shoulder surrogate is added like the one developed in §5.4. This would also be straightforward and inexpensive to integrate into a test house.

2. Pass limit of > 100 g (4900 N):
    a. The work completed to calculate the limit is displayed in §6.4.2. Cut and Abrasion injuries caused by blunt force impacts rather than stud impacts were of most importance. Padding's force attenuation properties were used as a measure of its ability to protect from Cut and Abrasion injuries.
    b. Using the testing completed in §6.4.2, the following steps summarise how the > 100 g (4900 N) was calculated:
        i. A reduction in force of 61% is seen between a cut/ tear – padding and a cut/ tear - no padding.
        ii. Padding should not attenuate more force than this limit.
        iii. Current padding is attenuating on average 60.5 % of the impact force in the Regulation 12 set up (9.8 J), the average peak impact force is 7710 N (157 g).
        iv. If the padding was to attenuate 61% of the impact force, the peak impact force would be 7628 N (156 g).
        v. Setting a 100 g (4900 N) pass mark where padding could not attenuate over 75% of the impact force would be appropriate for the following reasons:

1. Setting the limit to 75% attenuation would provide a safety factor of ~ 1.5 [236] (156 g / 100 g). This would ensure padding protects from Cuts and Abrasions.
2. Under this new regulation, current manufacturers' padding would still pass, meaning it would not disrupt product ranges.
3. The 100 g (4900 N) limit conveniently provides a good round number that is good for ease of communication and test house use.
4. Measurement of peak force using a load cell was recommended as this creates less noise and provides more accurate results than an accelerometer.
5. Padding's force attenuation properties were used as a measure of its ability to protect from Cut and Abrasion injuries. Therefore, the continued use of a steel rigid

anvil, rather than the anatomical or simplified surrogates presented in the thesis would provide a more repeatable, replicable and cheaper impact surrogate for testing.

7.3 Recommendation 2: Zones of Coverage

The current regulation states:

1. 'All body padding must comply with thickness and density requirements.' (Section 6.3.1 of the regulation).
2. Only padding in the shoulder region is tested for impact attenuation performance.' (Section 7.3.1 of the regulation).

It is recommended that all body padding should be subjected to all tests, and hence must conform to all requirements. This is because, at present, padding located outside the shoulder area could offer more impact protection than the defined limit for shoulder padding. Testing all paddings present on the clothing would ensure all locations conform to requirements.

7.4 Recommendation 3: Construction of Padding Material

The current regulation states:

'Padding materials must be homogeneous. Foam padding of sandwich construction is not allowed.' (Section 6.2.2 of the regulation).

It is recommended that non-homogeneous /sandwich construction is allowed. However, compliance testing of the padding in both directions to ensure no extra protection to the wearer should be completed. This is because homogeneity is ambiguous, as any moulding, segmenting, and shaping of homogenous padding material during manufacturing can give an inhomogeneous product. Moulded and shaped products have been approved before and this would allow for further non-homogeneous products to be approved.

7.5 Further Recommendations That Are Not Specific to Regulation 12

7.5.1 Stud Testing Protocols

§6.5 assesses padded rugby clothing's ability to protect from stud-induced injuries. Both the test methods outlined (stud impacts & stud raking), could be used to assess other forms of protection in Rugby Union like headgear and breast padding. Other forms of sports PPE that protect from stud injuries like shin pads in football could also be integrated into the testing protocols.

The testing protocols described in §6.5 were not recommended to be included in Regulation 12. However, these test protocols could be used to assess the ability of future products to protect from stud-induced injuries. This could be of keen interest both commercially, and to World Rugby™.

7.5.2 Testing of Other Rugby Equipment

The process of developing biofidelic human tissue surrogates and integrating them into an impact testing setup for the assessment of shoulder padding can be easily replicated for other forms of protection. In Rugby Union, the performance of headgear or breast padding could be assessed by following similar test protocols. With head injury in rugby being a concern; and female rugby growing rapidly, the use of biofidelic impact surrogates to assess the performance of these products should be considered.

The testing protocols could also be adapted for the assessment of padding in other collision sports like American Football or Ice Hockey. Additionally, other industries further afield, like the development of PPE in the ballistics industry could also benefit from the work completed in this thesis.

CHAPTER 8 - DISCUSSION, CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE WORK

8.1 Chapter Overview

In chapter 1, an overall research aim and research questions specific to WR and the individual project were set out. Each of these questions has been reviewed where appropriate. The limitations of the research have been addressed and subsequent future work recommended. How the work is novel and its contribution to knowledge is also outlined.

8.2 Discussion

No prior research had been published where a multi-layer shoulder surrogate had been developed. The development of this was a key requirement in addressing World Rugby's™ research questions and therefore recommending updated testing procedures for an improved Regulation 12 (padded clothing). The success of this project could be established through how comprehensively these research questions were addressed, the following looks to review this.

Q1: Can bespoke human tissue simulants be fabricated that give a consistent and biofidelic response to load?

Bespoke silicone formulations were developed for relaxed muscle tissue. The silicone exhibited a similar response to compressive load when compared with porcine muscle tissue. Compressive mechanical tests at quasi-static and dynamic strain rates were conducted to match the response of porcine muscle tissue. At quasi-static strain rates, the silicone simulant matched the porcine muscle tissue at strains over 0.5. An error of up to 100 % was seen at lower strains, potentially due to the elastic nature of the silicone and the porcine tissue's lack of tonicity. It was concluded this was not an issue due to the high strains seen in rugby impacts. At dynamic strain rates, the silicone exhibited a similar response to porcine tissue (> 40 percentage difference), however, this percentage difference increased as the strain increased.

The silicone formulation also exhibited a consistent compressive response to load, both when identical test samples were made and when they were tested 30 days apart. The maximum difference from the mean was around 30 %. Because of this, the silicone muscle simulant developed could be used to fabricate a repeatable shoulder surrogate.

Q2: Can a human shoulder surrogate with representative anatomies and geometries be fabricated feasibly in a repeatable manner?

In §5.5 the development of an anatomical shoulder surrogate is presented. The surrogate embodies some anatomical simplifications, notably, it only has two layers, a bone, and soft tissue (relaxed muscle) layer. Other anatomies that may modify the surrogate's impact response like tendons and fascia are also left out. A shoulder surrogate that represents the human's external and bone geometries were developed. This two-layer approach was deemed successful because a large amount of the soft tissue in the shoulder region is muscle. The approach also meant the surrogate could be fabricated feasibly at a low cost, in a repeatable manner. The surrogate provided a good improvement on the rigid steel anvil in the Regulation 12 (padded clothing) test and a greater understanding of padded rugby clothing's performance could be achieved.

Q3: Can a durable, biofidelic human shoulder surrogate be developed for repeatable and affordable use in a test house?

In §5.4 a simplified shoulder surrogate was developed, the reason for this was to aid repeatable testing of padded clothing's ability to protect from Cuts and Lacerations. The simplified nature of the surrogate enabled a multi-layer surrogate which included a skin layer.

The surrogate embodied a uniformed half-cylinder design with layer thicknesses consistent throughout. These simplifications in geometries are common practice in the sports industry when developing human surrogates [237]. In the development of this surrogate, the simplifications enabled affordable fabrication and repeatable testing while also keeping its biofidelity to be used to assess padded clothing's ability to protect from Cuts and Lacerations. The surrogate's repeatability was measured both in §5.4.4, at quasi-static loading rates and §6.3.2, at dynamic impact loads with percentage difference not exceeding 20 %.

Q4: Can human shoulder surrogates be integrated into an impact testing set up in order to reconstruct specific injurious scenarios relevant to evaluations of padded rugby clothing?

In §6 both a simplified and an anatomical shoulder surrogate have been integrated into an impact drop tower set up to understand padded rugby clothing's ability to protect from Cut and Laceration injuries. Padded clothing's force attenuative properties were the main method of measurement, while video footage and observation also provided a good measure of the padding's performance.

Cut and Laceration injuries caused both by blunt force (bony parts of rugby players' bodies) and by rugby players' studs were reconstructed using the simplified surrogate while a better understanding of both shoulder padding and the shoulder's response to impact was achieved with the anatomical surrogate. The project also shows that human shoulder surrogates could be integrated into other impact testing setups for use in other applications, for example, the assessment of American football shoulder padding or backstraps on rucksacks.

8.3 Limitations and Recommendations for Future Work

The development of human shoulder surrogates to assess the performance of padded rugby clothing and in turn shoulder injury biomechanics has not been previously investigated. It was therefore often not possible to compare the results of this project with other research. The main limitations impacting the work are described below. How these limitations guide recommendations for future work are also detailed.

8.3.1 Anatomical Assessments of the Human Shoulder

The data set used for both the internal and external assessments of rugby players' shoulders was low (< 10). The median participants' geometries were then used to guide the development of the shoulder surrogates. It is recommended that a larger data set should be obtained, if this is achieved, geometrical differences in BMI and playing position could be assessed. As well as this, the data could be used to compare shoulder anatomies to the general public, this would be useful in both research and industry.

Soft tissue layer thicknesses were examined using ultrasound. This is a fast, non-destructive technique for measuring the thickness of a material from one side. However, many factors affect its accuracy including proper instrument calibration, uniformity of material, sound velocity, sound attenuation and scattering, surface roughness, curvature, poor sound coupling, and backwall non-parallelism. The project used MRI data from literature but did not obtain any novel data from rugby players. If MRI data were obtained in the future, more accurate layer thickness measurements could be obtained. Comparing MRI data to scans from the general public would also provide valuable information for research and industry.

Assessments of the shoulder were only made in an anatomical position (arm relaxed by the participant's side) for consistency. In Rugby Union, tackles are often made with the arm being outstretched. This could alter the geometries of the shoulder. Future work should look to explore how the position of the arm can affect both external and internal shoulder anatomy.

8.3.2 Characterisation of Human Tissues

The characterisation of human tissue presents a key area of future work both in this project and in general. The current project examined the mechanical properties of porcine tissue in vitro. This poses two key issues, animal sources like porcine tissue, although provide a good representation of human tissues, will not exactly reproduce human mechanical response. Balaraman [108] found that human muscle tissue displays a slightly stiffer response to porcine muscle tissue. Furthermore, organic tissues in vitro, are likely to display a different mechanical response to the same tissue in vivo, due to the absence of physiological and neurological activity. Ethical issues make it difficult to test on in vivo human tissues, however further work in this research field would allow for the development of more mechanically accurate simulant materials.

The muscle tissue simulant materials developed in this project represented relaxed muscle. The effects of muscle contraction were never considered. Seminati [61] found that the Trapezius muscle does contract when making a tackle in rugby. It was chosen not to consider this due to the lack of research in this field and the varying levels of muscle contraction and biomechanics of a rugby tackle. Future research should consider the effect muscle contraction has on the mechanical properties of both muscle tissue and any of the tissues in the system.

Although the project did characterise porcine muscle tissues at dynamic strain rates, it mainly focussed on quasi-static strain rates. Organic tissues will exhibit a strain rate dependent response. Coupled with this, sports impacts generally exhibit dynamic strain rate conditions. Characterisation of porcine tissue through a detailed range of strain rates would add important information to the research field.

8.3.3 Human Tissue Simulant Materials

The silicone material developed in this project only represented that of muscle tissue. Further work should look to develop silicone formulations that match both skin and adipose tissues. The simulant muscle material developed looked to match the compressive isotropic properties of its human counterpart, this due to the loading direction of rugby impacts. Future research could look to consider the anisotropic behaviours of human tissues to improve the simulant materials.

Chamois leather was used to represent skin tissue due to its use in previous sports impact surrogates and its similar penetration properties. It also posed a cheap alternative to developing a new silicone formulation. This posed an issue when developing the anatomical shoulder surrogate as it could not be moulded around its complex geometries. Further work should look to develop a silicone skin simulant with representative mechanical properties to enable a three-layer anatomical surrogate.

8.3.4 Anatomical and Geometrical Representation of Human Shoulder Surrogates

Both the simplified and anatomical surrogate had many simplifications with reasoning for this mentioned in §5. Improvements in scanning techniques have made it possible to view many anatomies in the body, tendons, or fascia for example. These tissues may influence the mechanical response to impact. As well as the addition of a skin and fat layer, these other tissues should be considered when developing future anatomically accurate shoulder surrogates if technology and cost allow.

Full validation of the shoulder surrogates using human participants as planned in the project workflow (Figure 3.14) was also not completed. This was due to two reasons; ethical restrictions meant only low load testing that would not harm the participant could potentially be completed, this becomes an issue when developing impact surrogates. Secondly, the Covid-19 pandemic meant testing on human participants could not be completed. Future work, where the shoulder of humans are subjected to low load indentation tests to characterise its mechanical response should be completed.

8.3.5 Impact Testing of Padded Clothing

Although the work completed looked to match the simulant material's mechanical response to their human counterpart, to fully validate the impact test, it should be compared with living humans. Putting a living humans' shoulder into an impact test setup could be complicated due to ethical concerns and the addition of instrumentation could be challenging. However, it may be possible to conduct testing at low impact intensities with the addition of pressure sensors between the padding and human skin.

The surrogate used for impact testing was fixed to a rigid plate. In a rugby impact, the shoulder could be moving in any direction, therefore changing the loading condition in the impact. Coupled with this, impacts in rugby are very variable with ever-changing impact intensities and locations. The varying number of impact conditions does however provide an opportunity to develop impact surrogates of differing geometries, a shoulder surrogate with a raised arm for example, and use this in an impact set up with varying loads and intensities.

The project focussed on evaluating padded clothing's ability to prevent Cuts, Lacerations, and Abrasions through the assessment of damage to the chamois leather skin. Although an effective method to assess this, further work to relate these injuries to measurable mechanical response phenomena (i.e. magnitude of stress and strain) is needed. Padded clothing's ability to prevent other common impact injuries like Contusions or fractures was not assessed. The addition of pressure sensors embedded into the shoulder surrogate and further work into the mechanical response phenomena that cause these injuries should be completed to fully understand the injury preventative abilities of padded rugby clothing. These techniques could also be used in other sports and research fields like the ballistics industry.

8.4 Contributions to Knowledge

This project provides an original contribution to existing literature in Rugby Union, sports impact testing protocols, and human impact surrogate development. Its main contributions to knowledge are detailed in Table 8.1.

Table 8.1 – Projects main contributions to knowledge.

Contributions to Knowledge	How is this Novel
Rugby players' perceptions and attitudes towards padded rugby clothing were identified. Results from this can be used to guide both research and commercial developments in the future.	No previous studies relating to players' perceptions and attitudes towards padded rugby clothing had been previously published.
The prevalence of less severe shoulder injuries like Lacerations and Contusions were identified. These had previously been underreported due to the injury definitions used in research.	Prevalence data on rugby injuries that do not cause 24 + hours time loss was discovered.
The external and internal shoulder geometries from a data set of rugby players were identified. This data can be used to inform future surrogate design or be used for other means, like clothing design.	There is currently no published research that examines the geometries of rugby players' anatomies using technologies like 3D scanning and Ultrasound.
The compressive properties of porcine tissue at quasi-static and dynamic loading rates were measured. This data adds to a variable field of organic tissue data and can be used to guide organic muscle tissue simulant development.	Very few studies have looked to characterise the quasi-static and dynamic compressive response of organic muscle tissue
A commercially available silicone formulation was developed as a relaxed muscle simulant. This can be used in human surrogate development in the future.	The work completed presents a bespoke relaxed muscle simulant that has not been previously developed before.
The development and fabrication procedures of a simplified shoulder surrogate have been	A human shoulder impact surrogate has not been developed previously.

presented. These can be replicated and used for future shoulder injury biomechanics research.	
The development and fabrication procedures of an anatomical shoulder surrogate have been presented. These can be replicated and used for future shoulder injury biomechanics research.	A human shoulder impact surrogate has not been developed previously.
Mechanical test methods for the assessment of padded rugby clothing's performance were established. This can be used to test future designs of padding or used in other sports and industries.	No published research has established impact test protocols using biofidelic shoulder surrogates is available to date.
The research quantitatively evaluated padded clothing's ability to protect from Cut, Laceration, and Abrasion injuries. These results can be used to guide the future design of padded rugby clothing.	Although some research has tested the force attenuation characteristics of padded rugby clothing, no research has looked to quantify its protective abilities to specific injuries.
The research helped to guide recommendations made to WR concerning the standards and test protocols related to padded rugby clothing.	WR will now have an updated Regulation 12 (padded clothing) that is backed up by science.

9. APPENDICIES

9.1

Appendix A: Images of each participant's body scan

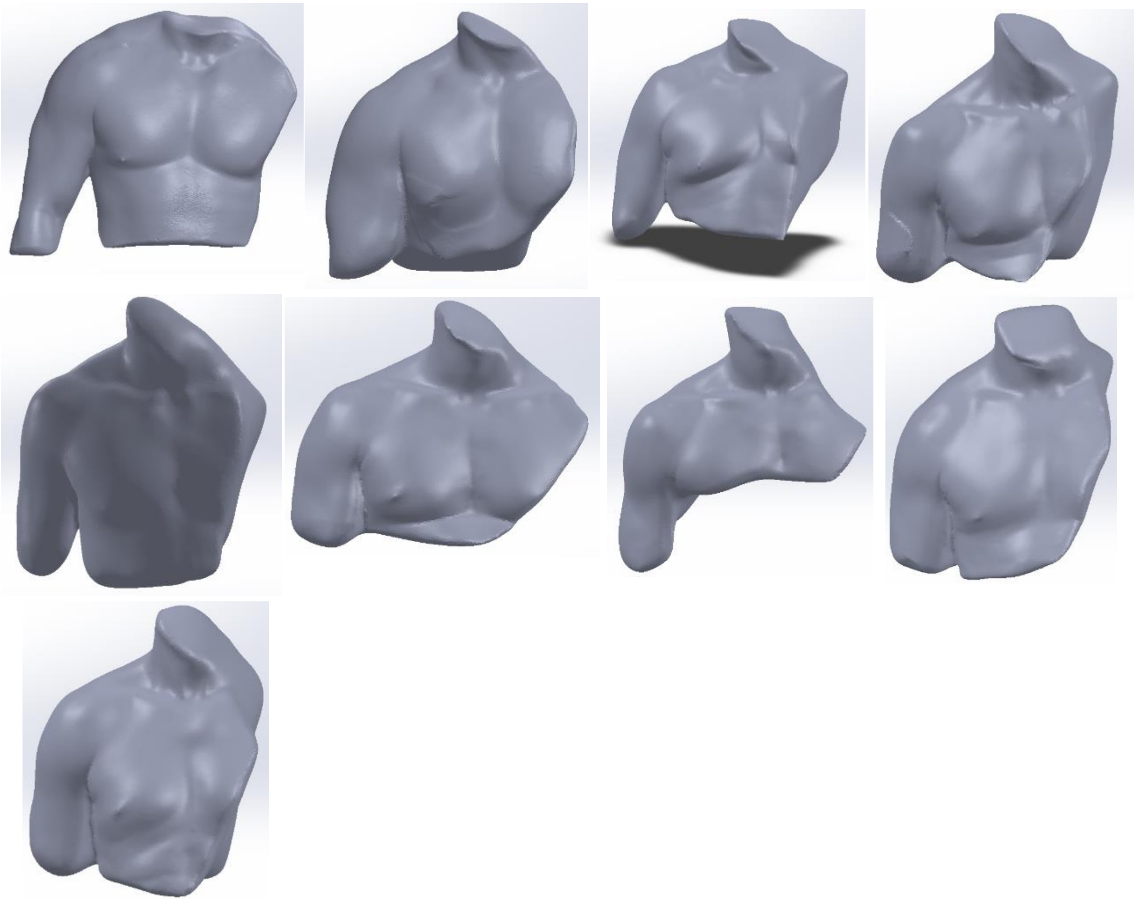


Figure 9.1 – CAD drawings of each participant's shoulder scan.

9.2

Appendix B: Simplified Surrogate Assembly Drawing

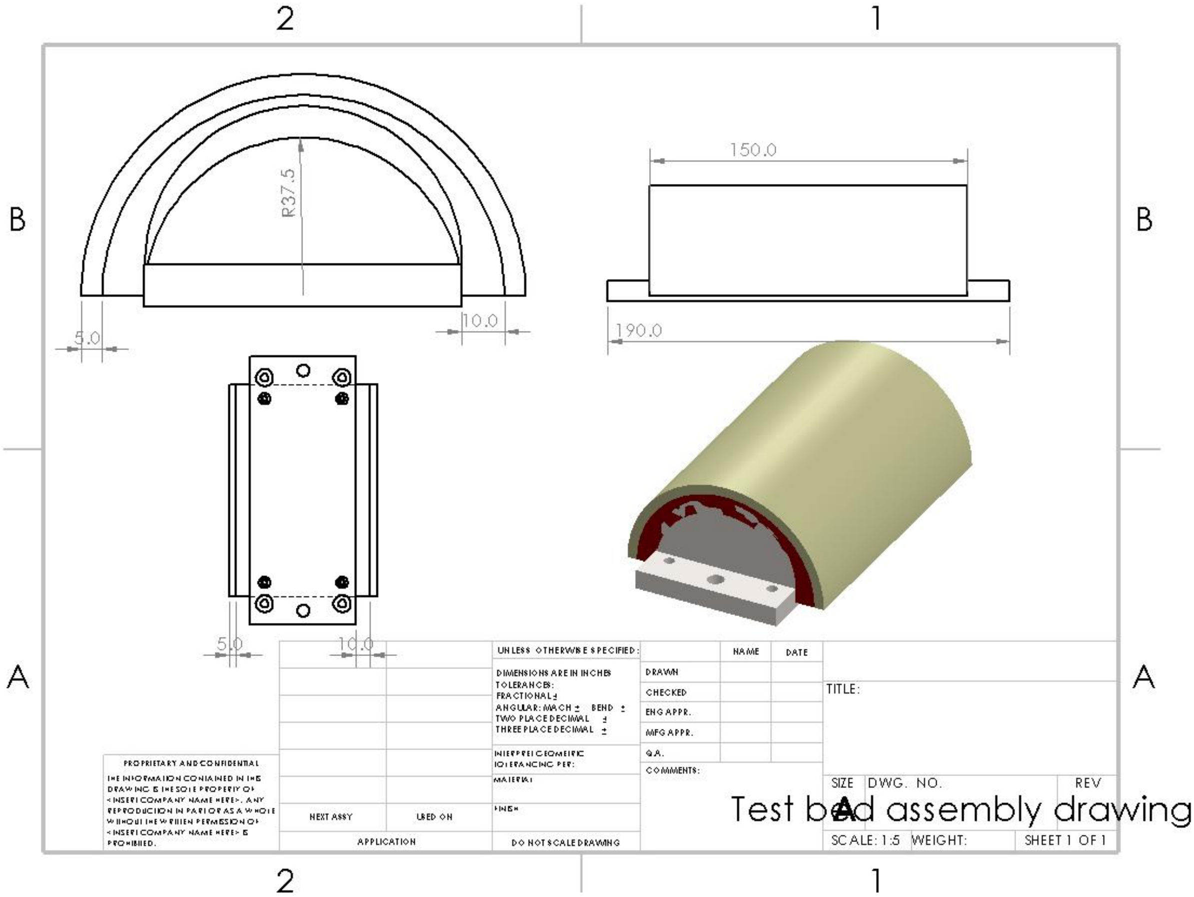


Figure 9.2 – Assembly drawing of the simplified surrogate.

9.3

Appendix C: CAD Drawing of Intended Drop Rig at UoS

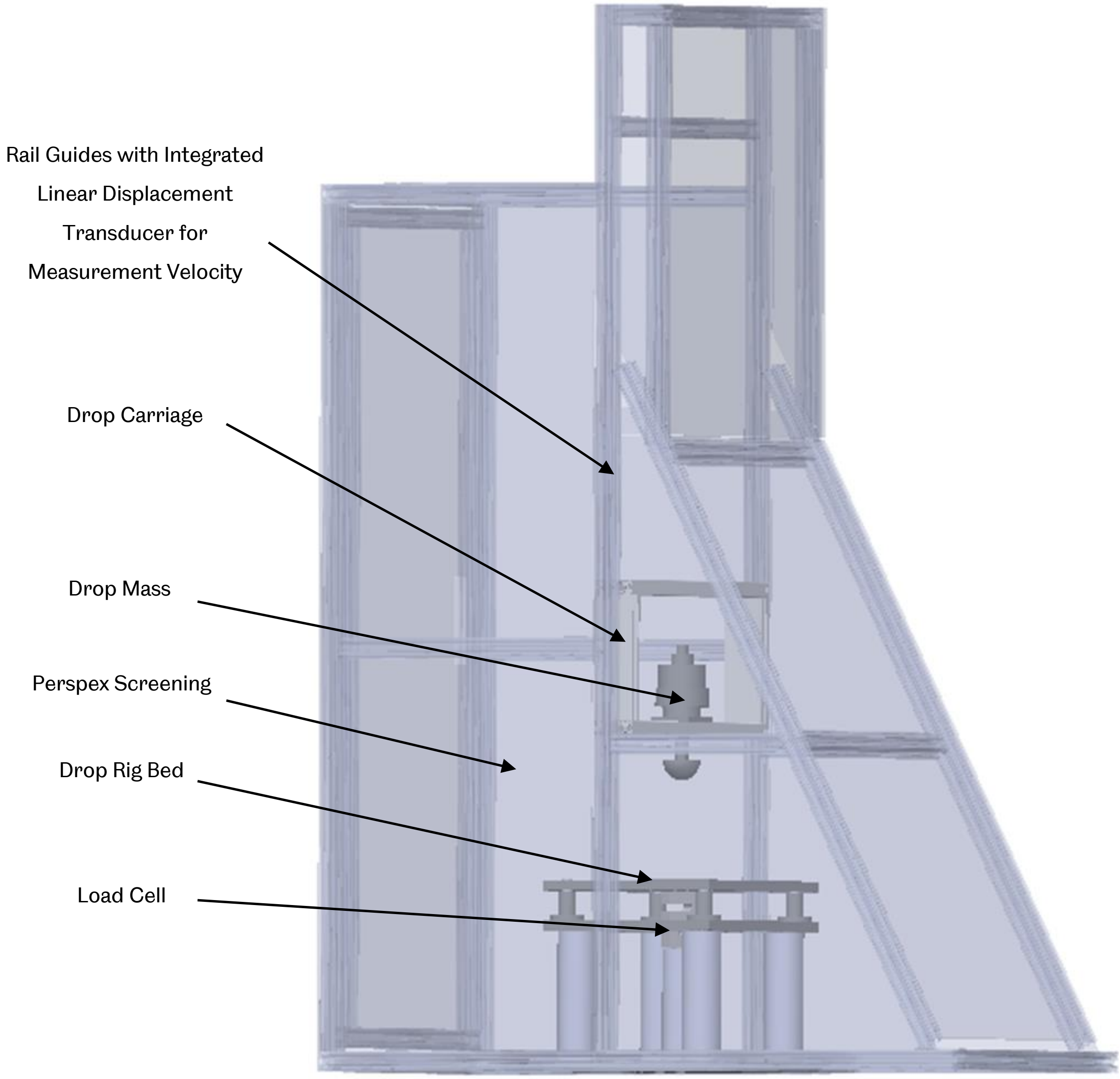


Figure 9.3 – Labelled CAD drawing of Drop Rig planned for Development at UoS.

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