CARDIFF UNIVERSITY

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AN INVESTIGATION TO ESTABLISH HOW THE EVOLUTION OF RUGBY INFLUENCES THE RISK OF SPINAL INJURY DURING SCRUMMAGING

By

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THESIS SUMMARY

The rugby scrum results in a large number of injuries to the players of the front row, particularly the hooker. Front row players have been known to suffer from acute and chronic injuries of the cervical spine as well as low back pain.

The principal goal of this research was to develop a method to measure spinal biomechanics of the hooker's role during rugby scrummaging and use this method to address the question of whether recent changes in rugby affect the risk on hooker spinal injury. In recent years, a number of changes have occurred in rugby which may have an effect in injury risk, however, little is currently known about the effect of these changes. This was accomplished through three experimental stages. Firstly, a review of kinematic measurement techniques was undertaken in order to determine the most feasible measurement technique. Inertial sensors were chosen and validated for orientation output against high precision digital encoders with high levels of concordance for each axis of each sensor (>0.95). In addition to this, a method of using electromyography to predict muscle force production was investigated. Determined force and recorded force were found to be insignificantly different ($p>0.05$) across the 12 participants investigated. Having proved this to be a feasible method, it was put into practise in-field. Inertial sensor technology was combined with a laboratory tested force-EMG correlation to assess spinal biomechanics during live, contested, training scrums for an initial sample of 9 rugby union hookers. No significant differences ($p>0.05$) were observed for peak kinematic variables or EMG data.

The second study assessed whether a change in playing surface affects hooker spinal kinematics by evaluating key variables such as peak range of motion and angular velocity as there has been a recent shift towards the use of artificial surfaces. Twenty-two participants took part in this study with 11 participants in each group. The groups were not significantly different ($p>0.05$) in terms of anthropometric and background information. The results of this study indicated that key kinematic variables did not differ significantly ($p>0.05$) between playing surfaces. There was, however, a large effect ($d>0.8$) for certain peak angular velocity measurements of the thoracic region.

The final study investigated how different engagement techniques affected hooker spinal biomechanics since a recent law change was introduced governing the scrum. These techniques included machine scrummaging and live scrummaging of two different engagement sequences. Twenty-nine participants took part in the live-vs-live comparison with 14 of those taking part in all three experimental conditions. The results of this final study indicate significant biomechanical differences ($p<0.05$) between machine and live scrummaging indicating that machine scrummaging is a much more constrained environment. Live scrummaging of the two different sequences did not yield any significant differences ($p>0.05$) for both kinematic and muscle activity/determined force data indicating that the sequences do not affect hooker spinal biomechanics.

These results suggest that the recent changes in rugby do not significantly affect the risk of spinal injury of the rugby union hooker.

ACKNOWLEDGEMENTS

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LIST OF THESIS PUBLICATIONS

The original research articles published as part of this research are presented in the following articles:

1. **Swaminathan R,** Williams JM, Jones MD, Theobald PS. 2015. The prediction of neck extensor force using surface electromyography. *Journal of Back and Musculoskeletal Rehabilitation.* (Published Online)
2. **Swaminathan R,** Williams JM, Jones MD, Theobald PS. 2015. A Kinematic Analysis of the Spine during Rugby Scrummaging on Natural and Synthetic Turfs. *Journal of Sports Sciences.* 34(11), pp. 1058-1066.
3. **Swaminathan R,** Williams JM, Jones MD, Theobald PS. 2016. Does the New Rugby Union Scrum Technique Positively Influence the Hooker's *In Situ* Spinal Kinematics. *BMJ Open Sport and Exercise Medicine.* (Published Online)

The conference proceedings published as part of this research are presented in the following articles:

1. **Swaminathan R**, Williams JM, Jones MD, Theobald PS. 2014. An Assessment of the New Scrum Laws on the Spinal Kinematics of the Rugby Union Hooker. *Proceedings of the 7th World Congress of Biomechanics, Boston, USA*.
2. **Swaminathan R**, Jones MD, Williams JM, Theobald PS. 2014. Electromyography of the Cervical Spine during Machine and Live Scrummaging. *BASES Conference, Portsmouth, UK*.
3. **Swaminathan R**, Jones MD, Williams JM, Theobald PS. 2014. An Assessment of the Effect of the New Rugby Union Engagement Laws on the Lumbar Spine Kinematics of the Hooker. *BASES Conference, Portsmouth, UK*.
4. **Swaminathan R**, Jones, MD, Williams JM, Palmer T, Theobald PS. 2013. The Range of Motion of a Rugby Union Hooker during 'Machine-Based' Scrummaging. *BASES Conference, Cardiff, UK*.
5. **Swaminathan R**, Jones, MD, Williams JM, Palmer T, Theobald PS. 2013. The Range of Motion of a Rugby Union Hooker during Live Scrummaging. *BASES Conference, Cardiff, UK*.

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LIST OF ABBREVIATIONS

ACC	Accident Compensation Corporation
CBS	Crouch-bind-set engagement sequence
CES	Cervical erector spinae
CTPE	Crouch-touch-pause-engage engagement sequence
CTS	Crouch-touch-set
C-spine	Cervical spine
EMG	Electromyography
HSV	High speed video
IRB	International Rugby Board
LBP	Low back pain
L-spine	Lumbar spine
NZRU	New Zealand Rugby Union
PB	Crouch-touch-set with pre-binding of props (similar to CBS)
RFU	Rugby Football Union
ROM	Range of motion
SARU	South African Rugby Union

1. Introduction

Rugby union is a contact sport that involves periods of intense physical activity and then periods of relatively low activity such as walking and jogging. High intensity activities include rugby-specific events such as tackling, rucking, mauling, lineouts and scrummaging as well as non-specific activities such as sprinting (Roberts et al. 2008). Rugby has a relatively high injury incidence (Brooks, Fuller, Kemp, & Reddin, 2005; Brooks & Kemp, 2008) compared to other professional sports. Given the high incidence of the sport, there has been research focus on certain aspects of the game such as the tackle and the scrum. With regards to biomechanics, the scrum has been a particular area of focus as it is relatively controlled in comparison to the tackle. The scrum is associated with a high risk of injury (Fuller, Brooks, Cancea, Hall, & Kemp, 2007; Roberts, Trewartha, England, & Stokes, 2014; Taylor, Kemp, Trewartha, & Stokes, 2014) and has resulted in both acute (Secin et al. 1999; Wetzler et al. 1998) and chronic (Broughton 1993; Scher 1990b; Berge et al. 1999) injuries.

The purpose of the scrum is to restart the game after certain minor infringements in a quick, safe and fair manner (IRB 2013). The scrum consists of two opposing 'packs' each consisting of 8 players arranged in 3 rows. The interlocking of the heads' of those in the front row creates a tunnel into which the ball is fed (Figure 1).

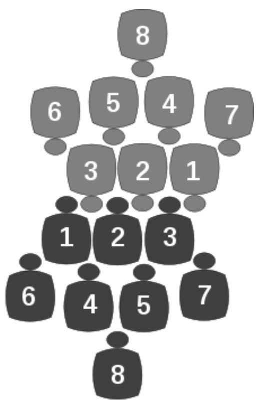

Figure 1 - Relative positions of players during a rugby scrum. The numbers represent those typically associated with a particular playing position. Front row players are typically numbered 1 (loose-head props), 2 (hookers) and 3 (tight-head props)

Pack engagement is controlled through the verbal direction of the referee and, in recent years, has been frequently evaluated to ensure that the scrum best meets the above objectives. These verbal sequences used have included 'crouch and hold-engage', 'crouch-touch-set' and 'crouch-touch-pause-engage'. As of the 2013-14 playing season, this has yet again been changed to 'crouch-bind-set' and this change has been associated with a reduction in engagement force and key kinematic variables, such as engagement velocity and accelerations of upper spine landmarks, by up to 20% (Cazzola et al. 2015).

1. 1. Statement of the Problem

Spinal injuries in sport are a serious problem and can result in long-term social and economic impacts to the individual as well as the family and wider society. For a single spinal cord injury that results in partial or complete paralysis it is estimated that, on average, lifetime costs will be between $2-3 million (Quarrie et al. 2002; Dennison et al. 2012). These types of injuries are relatively uncommon in sport but certain sports carry a greater risk than others because of their highly physical nature. Such sports include American Football (Torg and Ramsey-Emrhein 1997; Torg et al. 2002), ice hockey (Tator 1987; Carll et al. 2010), combat sports such as martial arts (Kocchar et al. 2005) and rugby (Secin et al. 1999; Quarrie et al. 2002; Boran et al. 2011).

Rugby carries a particular risk of injury to the spine. Despite the intention of the scrum to be a safe and stable platform to restart the game, it is the cause of 5.6-12.6% of all injuries to forward players and is hence one of the most injurious events within the game (Brooks, Fuller and Kemp 2005; Fuller, Brooks, Cancea, et al. 2007; Roux et al. 1987; Schick et al. 2008; Taylor et al. 2014; Taylor et al. 2011). Scrummaging causes a variety of injuries including those to the lower leg musculature (up to 54%) and shoulder injuries (up to 66%) (Brooks, Fuller and Kemp 2005; Brooks and Kemp 2011). Of more concern, however, is the proportion of spinal injuries that result from the scrum. This has been quoted to be between 37-61% of all scrummaging injuries (Armour et al. 1997; Bohu et al. 2009; Dunn and van der Spuy 2010; Secin et al. 1999; Kew et al. 1991; Wetzler et al. 1998; Wetzler et al. 1996). This large proportion of spinal injuries sustained during the scrum are largely to the cervical (Wetzler et al. 1998; Trewartha et al. 2015; O'Brien 1996; Scher 1990b) and lumbar (Fuller, Brooks and Kemp 2007; Castinel et al. 2007) regions. With specific regards to the cervical spine, injuries can be largely categorised into acute and chronic injuries. Both these types of injuries are well documented to result from the scrum but empirical evidence demonstrating that scrummaging may be a direct cause of these types of injuries is currently lacking.

Previous research has focussed on the machine scenario (Milburn 1990; Quarrie and Wilson 2000; Preatoni et al. 2013) until more recently where some studies have investigated the live scrummaging scenario (Cazzola et al. 2015; Cazzola et al. 2014a). Even with the live scrummaging scenario all previous research has focussed on the characterisation of shoulder impact and sustained forces during this scenario with little direct attention being given to spinal loading. Furthermore, previous research has generally defined numerous body segments in two dimensions (Preatoni, Wallbaum, et al. 2012; Sayers et al. 2009; Rodano and Pedotti 1988) but no studies have previously attempted to define three dimensional motion of any segment. Loading of the shoulder complex and two-dimensional kinematic analysis give researchers, to a certain extent, a relatively limited view of this unique biomechanical loading scenario. Thus, the aim of this study was to attempt to characterise, to a certain extent, spinal loading and well as provide a comprehensive three-dimensional kinematic analysis of the spine.

1.2. Purpose of the Study

The overall purpose of the study was to establish how recent changes in rugby have affected the risk of spinal injury of the hooker. These changes include a shift towards artificial playing surfaces and the recent law change governing the scrum and the main two chapters of this thesis describe an investigation of these changes. Firstly, a valid method to analyse spinal biomechanics during the rugby scrum was determined through a series of controlled laboratory tests and preliminary in-field testing. Classical mechanics consists of two primary branches; kinematics and kinetics and the same is true for biomechanics. Thus, the ability to measure both branches concurrently will provide invaluable biomechanical data. The purpose of the first set of experiments was to determine whether the chosen kinematic measurement technique was suitable and accurate for kinematic data collection. The second set of experiments sought to validate a method to measure force production in the cervical spine. More specifically, it attempts to use electromyography (EMG) as a means of predicting cervical extensor force production as this would provide a method for measuring force production during live scrummaging. A relationship between force and EMG was developed and then EMG alone was used to predict muscle force production. A combination of these two methodologies was then used in the final sub-section of this chapter to measure biomechanical parameters of the spine of the hooker during machine and live scrummaging in a preliminary study and to identify any problems so these can be addressed for latter studies.

Having identified this opportunity for novel research, the thesis continues by investigating the effect of changing the playing surface on hooker spinal kinematics. In recent times, there has been a gradual shift towards using synthetic turf for rugby. Much previous research has concentrated on injury epidemiology and the shoe-surface interface but no study has previously investigated whether changing the playing surface has any effect on scrummaging biomechanics. This part of the thesis focussed on whether any change in key kinematic variables were present when participants scrummaged on natural turf or synthetic turf (3G) and how this relates to the risk of spinal injury.

The purpose of the final study of the thesis was to investigate whether different scrummaging conditions including machine scrummaging and live scrummaging, of two different engagement sequences, have any effect on hooker spinal biomechanics. The primary aim was to establish whether the recent change in scrummaging laws have had any effect on hooker spinal biomechanics. The secondary aim was to establish whether machine scrummaging provided an accurate platform on which to practise scrummaging from a biomechanical perspective.

The thesis presents results from the studies outlined above. It is hoped that the application of a new method to investigate spinal biomechanics will allow researchers to begin to gain an understanding as to why front row players are susceptible to spinal injuries. Furthermore, the collection of kinematic data for multiple spinal segments simultaneously means that either regional or segmental analysis can be performed. The knowledge gained from the analysis of spinal kinematics and, where possible, muscle activity developed our understanding of spinal injury mechanisms and why there is an onset of premature spinal degeneration in front row players. This knowledge will be invaluable in understanding underpinning injury mechanisms and if there is a possibility of reducing this for the player in question.

1.3. Organisation of Thesis

Chapter 1 provides an introduction to the problem, the purpose of the study and organisation of the thesis.

Chapter 2 provides a comprehensive review of the literature relating to kinematic measurement techniques, force-EMG relationships, rugby injury epidemiology and scrummaging biomechanics.

Chapter 3 present the methods used in the thesis across all studies.

Chapter 4 presents the findings of a series of laboratory studies validating the chosen kinematic measurement technique, using EMG to predict muscular force production and combining these methodologies for in-field testing.

Chapter 5 presents the findings of a study investigating the effect of playing surface on the spinal kinematics of the hooker.

Chapter 6 presents the findings of a study investigating the effect of different engagement conditions of hooker spinal biomechanics.

Chapter 7 presents a general discussion relating to the findings of this thesis.

Chapter 8 presents conclusions and recommendations for future work

2. Review of the Literature

The literature has been broken down into separate sections to cover a variety of topics. The topics include reviewing different kinematic measurement techniques, a review of the current understanding of the relationship between muscular force production and a review of rugby-related literature. The review of rugby-related literature includes injury epidemiology, injury mechanisms, playing surface literature and scrummaging biomechanics.

Firstly, a basic overview of spinal ROM terminology will be presented to allow for easier understanding of terms used during the thesis. Figure 2 shows the anatomical planes with the standing position being defined as neutral as well as the three basic motions away from the neutral position. Motion in the sagittal plane is characterised as flexion-extension (below – left side), motion in the coronal plane is lateral bending (below – middle) and motion in the transverse plane is rotation (below – right side). In the sagittal plane, bending forward from the anatomical position is defined as flexion and bending backward is defined as extension. The other motions, defined from the neutral position, are left and right lateral bending and left and right rotation.

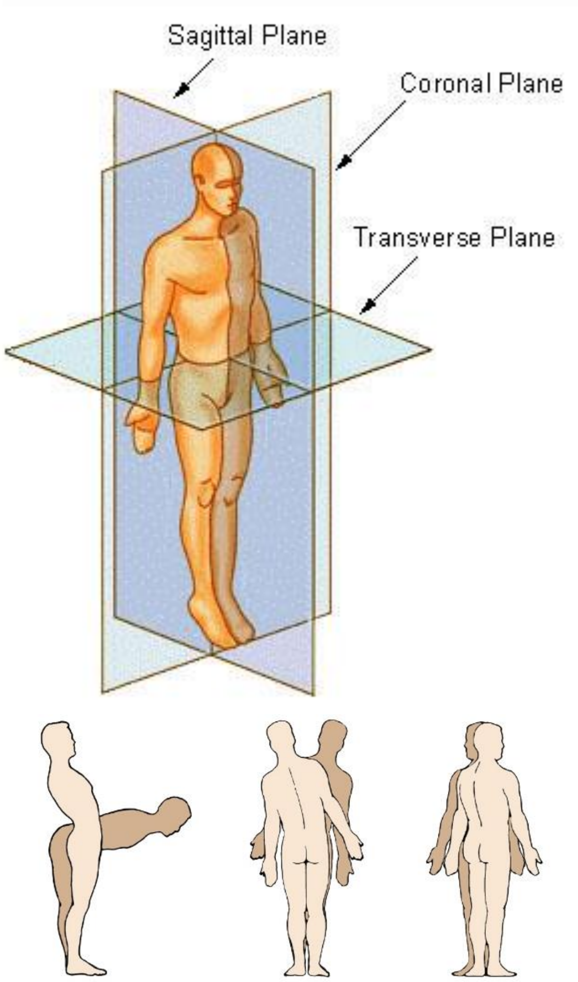

Figure 2 - Anatomical Planes (Top), Spinal Motion (Bottom); Bottom left – flexion-extension, bottom middle – lateral bending, bottom right – rotation

2.1. Review of Kinematic Measurement Techniques Literature

There are numerous different methods by which kinematic data can be collected ranging from relatively simple devices such as the flexicurve and inclinometer to more complex methods such as electromagnetic and opto-electronic systems. These techniques are reviewed with regards to their accuracy and suitability for the application in question; an assessment of spinal kinematics during rugby scrummaging. Each of these systems has the capability to measure range of motion (ROM) but certain systems can also measure the time-history of motion such as opto-electronic systems. Furthermore, it is possible to measure curvature through the use of multiple markers when using a system such as an opto-electronic one although there are limitations on how detailed this curvature can be measured.

Kinematic measurement techniques can largely be separated into two main categories; clinic-based systems and laboratory-based systems. Clinic based systems are generally simple to use but can only provide limited kinematic data. Laboratory based systems can provide much more detailed kinematic data but are often very costly, have environmental constraints and involve complex data processing techniques. Thus, there are advantages and disadvantages of each of these categories of systems.

2.1.1. Clinic Based Systems

Within the clinic, three main systems are used; the inclinometer, the electrogoniometer and the flexicurve. An inclinometer is a device that is used for measuring angle with respect to gravity and thus, can only measure in two dimensions. A goniometer is a device that is used for measuring joint angles and is often used in physical therapy to assess progress in a rehabilitation programme. It has two arms joined at a pivot point which can be aligned with a certain parts of the body, the thigh and shank for example, to measure knee ROM. An electrogoniometer is the same except that instead of manually reading off the scale, an electrical output is generated that is often transferred to a computer. Finally, the flexicurve is a modified draftman's curve which can be moulded over the spine and then the curvature can be traced to paper. Modern day flexicurves, however, electrically plot this curvature onto a computer. From this, ROM and curvature calculations can be performed.

All of these methods measure motion on the skin surface and they are not attached to the underlying bone. Thus, skin motion artefact is a significant problem and is a problem associated with many motion analysis techniques. It is an assumption for these techniques that the motion of the marker on the skin is representative of the underlying bone which is in fact not the case. There is a margin of error that is introduced by making this assumption but it is an assumption that is widely accepted as the only way of measuring true motion of the underlying bone using these methods would be to use invasive markers.

The inclinometer has been shown to have good intra-rater reliability for total lumbar ROM and lumbar flexion when compared to radiographs. Correlation coefficients were 0.94 ($p<0.001$) for total lumbar ROM and 0.88 ($p<0.001$) for lumbar flexion. The intra-rater reliability for extension, however, was poor with a correlation coefficient of 0.42 ($p<0.05$). It should be noted, however, that this was only for two operators but was performed across 54 participants (Saur et al. 1996). Furthermore, although correlation was extremely high, Lin's coefficient of concordance may have been a better method to determine the reliability of the method as it provides a value of how closely the two techniques measure the same variable. This method has been shown to be reliable for certain movements, but there is a high level of error associated with this technique (Bierma-Zeinstra et al. 2001). Furthermore, it can only provide data in a single dimension at any one time and can only be used for static measurements and thus, is not a viable option for use during scrummaging.

A significant improvement on the previous system is the electrogoniometer as it can provide dynamic measurements and data in two planes simultaneously. It is highly reliable in two planes simultaneously with an accuracy of less than ±1° (Feipel et al. 1999; Christensen 1999). The electrogoniometer provides an accurate and dynamic measurement system for the analysis of spinal kinematics although the device is restricted to two-dimensional analysis. It would hinder binding during scrummaging and can only provide data in two planes simultaneously and is therefore, not a viable option.

Finally, with regards to clinic based systems, there is the flexicurve. This technique allows for the difference in orientation of two points and spinal curvature data to be collected (Burton 1987). The difference between these two measurements is that the former gives the researcher data relating to how much the angle has changed between two points. Curvature, however, gives the researcher a better idea of how these two points are linked; that is, whether they are in a straight line or there is some curvature. It can only provide data in the sagittal plane and only for static postures however, meaning no dynamic measurements or out-of-plane motion can be recorded. It is thus, not a method that can be used for rugby scrummaging.

Despite the simplicity and ease of use of clinic-based kinematic techniques, they do not provide the depth of analysis that is required and so these systems must be rejected. Laboratory systems, although much more complex, may provide a potential solution and these are reviewed in the following section.

2.1.2. Laboratory Based Systems

2.1.2.1. Electromagnetic Tracking Systems

Electromagnetic tracking systems locate small electromagnetic sensors in an electromagnetic field which has a known geometry. The advantage of such a system is that no line of sight is required but they can easily be distorted by nearby metal sources. This device is highly accurate with rotational resolution of 0.1˚ (Milne et al. 1996) and average errors of less than 1.5˚ for two different motions when compared against an opto-electronic system (Hassan et al. 2007).

The limitations of such a system are important to consider as they underline the key sources of error that are inherent within its use. With the generation of an electromagnetic field, there is an operational zone in which the sensors must be placed. To increase the operational zone, the power of the transmitter must be increased (Bull and Amis 1997). If the power is not increased and measurements are taken outside of this area, errors in both position and orientation increase (Schuler et al. 2005). This operational zone limits the device's use for activities that are more dynamic as the subject is constrained to stay within this zone. Another significant limitation of the system is the errors introduced in the presence of metal (Milne et al. 1996).

Electromagnetic tracking systems are accurate, relatively small, transportable, can measure three-dimensional motion and are non-invasive. Limitations to the size of the operational zone and the effect of ferrous metals, however, mean that it is not a viable system for rugby scrummaging.

2.1.2.2. Opto-Electronic/Video Based Systems

Opto-electronic systems detect certain forms of light and, in the case of biomechanics, this is often infrared light. Motion capture cameras emit infrared light which is reflected back by retroreflective markers and the cameras can then track the movement of the markers. This description is a basic system but there are more complex systems which involve markers emitting signals back to the cameras as opposed to passively reflecting the infrared light.

These systems are expensive, require a line of sight for the cameras and are often confined to a certain environment. The system can be moved but system setup and calibration is an extremely time consuming and difficult process. The greatest advantage of this type of system is its flexibility in capturing kinematic data. The landmarks being considered can be changed with relative ease with the inclusion or exclusion of reflective markers. Wong and Wong (2008) used motion analysis and inertial sensors to determine spinal posture. Motion analysis systems provided continuous data on spinal curvature. Although this is a useful method, the constraints regarding the capture environment, the difficulty of moving the system and line of sight problems make it an unsuitable option for the application in question.

This system offers a viable option to measure spinal kinematics in three dimensions and the time-history of movements. These systems are, however, extremely expensive, difficult to transport and data analysis is often complex. Along with the environmental constraints associated with such a system, it is not a viable option for use in scrummaging.

2.1.2.3. Inertial Sensors

A number of industries have used inertial sensors but in recent times they have been used more frequently in biomechanics. The term 'inertial sensor' technically refers to accelerometers and gyroscopes but more often than not, the term refers to a combination of tri-axial magnetometers, accelerometers and gyroscopes. The reason a combination is used is that each component has its limitations but the combination of them allows some of these limitations to be eliminated. Inertial sensors have previously been used in trunk posture monitoring (Wong and Wong 2008), activity recognition (Avci and Bosch 2010), elite sports monitoring (James 2006), the assessment of human movement (Theobald et al. 2012; Saber-Sheikh et al. 2010) and, recently, for kinematic measurements during rugby scrummaging (Cazzola et al. 2015). This sub-section evaluates the components individually as well as the combination of them to form inertial sensors.

Inertial sensors provide a useful and viable option of measuring dynamic spinal motion because they are relatively cheap and do not suffer from problems such as line of sight or a specific capture volume. They have been previously used in postural analysis

(Hansson et al. 2001; Wong and Wong 2008) and have proved to be a feasible method. Wong and Wong (2008) used three inertial sensors in their analysis and so the inclusion of multiple sensors in a string would be a useful extension to this technique. Sensors could be placed at minimal intervals along the length of the spine to create multiple spinal segments. This method would overcome some of the limitations outlined with previous analysis methods. Through this method, a dynamic measure of spinal motion could be obtained for multiple segments. A significant advantage of inertial sensors is that they do not have many environmental constraints and so can be used for in-field research. A disadvantage of using multiple sensors in a string is that the complexity of analysis would increase greatly.

Figure 3 shows the basic setup of a one-dimensional accelerometer. The piezoelectric material (sensing element) creates a charge when some force is applied to it. This force is produced by the inertia of the mass on top of the crystal as it is accelerated by some motion which is to be measured. Despite their name, accelerometers do not measure acceleration directly. They measure the force applied, through the sensing element, and acceleration can be calculated through Newton’s Second Law of Motion (F = m x a). It should be noted that this is only one type of accelerometer and there are other types that use capacitors as well as simple mechanical accelerometers but the way in which acceleration is calculated is the same; through Newton’s Second Law of Motion.

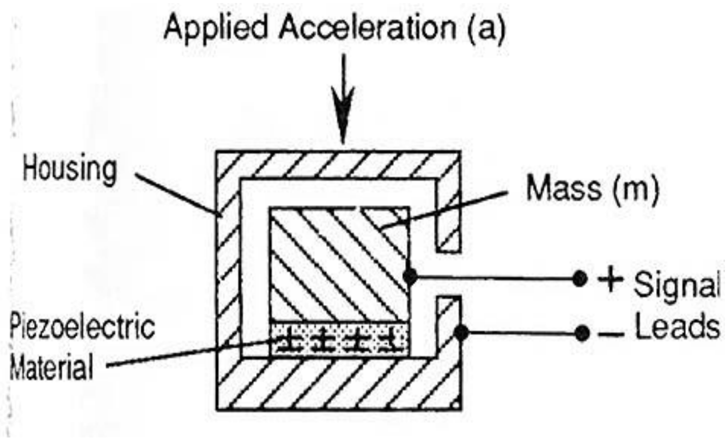

Figure 3 - Basic Accelerometer Setup (Image reproduced from www.pcb.com)

When static, an accelerometer can measure tilt with respect to gravity, acting as an inclinometer. Tri-axial accelerometers have a small angular error (1.3˚) and high reproducibility (0.2˚). As well as this, the angular noise of accelerometers is as little as 0.04˚

(Hansson et al. 2001). When compared to a motion analysis system, accelerometers demonstrated a precision of ≤1˚ for static calibration and an ICC value of 1.000. These findings show that this device is highly reliable (Wong and Wong 2008). Some uses for accelerometers in biomechanics are in posture analysis (Hansson et al. 2001; Wong and Wong 2008), the analysis of locomotive activities in the household (Oshima et al. 2010), measuring head movement in the assessment of neck pain (Jasiewicz et al. 2007) and assessing human movement (Saber-Sheikh et al. 2010). The biggest limitation of accelerometers is their inability to measure in three planes simultaneously. As a result of using gravity to measure angle of inclination, only two planes of motion can be analysed at any one time. Wong and Wong (2008) found that at a pre-tilted angle of ±80˚ the RMS error of the accelerometer was >1˚ because the accuracy of the device decreased as the sensing axis tends to the horizontal. Another limitation is that when measuring motion that is not constant in speed or direction, an angular error is introduced and thus the interpretation of the inclination data may not be valid (Hansson et al. 2001).

A gyroscope measures angular orientation through integrating a measured rotational rate of acceleration. It measures angular velocity based on the measurement of the Coriolis force. Figure 4 shows a diagram of the Coriolis effect. When a mass (m) is moving in the direction, x, and angular rotation is applied (Ω), the mass will experience a force in the direction of the arrow ($F_{Coriolis}$). The physical displacement resulting from this force can be measured from a capacitive sensing element, similar to an accelerometer. In gyroscopes, there are generally two masses that oscillate in opposite directions. When an angular velocity is applied the Coriolis force is generated in opposite directions which results in a change in capacitance. This change is capacitance is proportional to the angular velocity and this is converted into a voltage. Thus, from the change in capacitance, an angular velocity can be determined and then integrated to determine angular orientation.

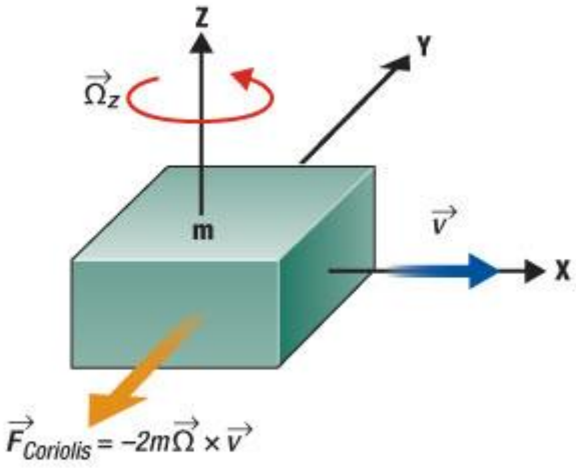

Figure 4 - The Coriolis Effect (reproduced from eletroiq.com)

This system has high reliability for anatomical movements with coefficient of multiple correlation (CMC) values ranging from 0.972 to 0.991 (Lee et al. 2003). Gyroscopes, however, suffer from integration drift which arises from the integration of the rotational velocities. This drift happens when static and can result in large errors after integration (Boonstra et al. 2006). Some solutions to this problem include Kalman filters (Lee et al. 2003), quaternion filtering (de Vries et al. 2009), and high pass filtering (Boonstra et al. 2006). High pass filtering can be problematic as it results in the loss of some gyroscopic data meaning an underestimation of the measured angle. Quaternion filtering also did not yield acceptable results but the Kalman filter did. A Kalman filter is a mathematical error model of an inertial sensor's characteristics and uses this to predict the error in orientation estimation which is then corrected. The filter, however, is based on two assumptions: apart from gravity, the averaged acceleration over a 10 second period is zero and the Earth's magnetic field is homogenous (de Vries et al. 2009). Thus, despite the Kalman filter's success in correcting gyroscopic drift, it has some major assumptions making it far from perfect. These are significant assumptions as if the inertial sensor is moving, acceleration over a 10 second period will not be zero. Also, the Earth's magnetic field is not completely homogenous and so this assumption may be considered acceptable although it is not completely true. A final potential solution to this problem of gyroscopic drift is the integration of an accelerometer to measure tilt in static conditions (Wong and Wong 2008). In this instance, an algorithm is developed to determine when the sensor is stationary and orientation data relative to gravity is used to correct for drift. In modern inertial sensors, a combination of these solutions is used to provide orientation data that is free of drift.

The final discrete component of inertial sensors are magnetometers. Although they are not ‘inertial’ in the way in which they function like accelerometers and gyroscopes, these devices are generally included in modern day inertial sensors to provide dynamic three-dimensional orientation data. Magnetometers are devices that measure magnetic fields and measure their magnetic flux density in three dimensions. The vector returned from the sensors describes the magnitude and direction of the magnetic field relative to the device. Through these 3D co-ordinates, it is possible to calculate the orientation of the device. When compared to an opto-electronic system, magnetometers provided a reliable method of orientation measurement. The average static error was 1.4° (SD = 0.4°) when magnetic distortion was introduced but this error increased to 2.6° during dynamic experiments (Roetenberg et al. 2005). Furthermore, the inclusion of magnetometers in inertial measurement systems has been shown to aid in the measurement of orientation (Farrell 2003).

Measures of repeated reliability for inertial sensors have previously been performed with regards to the cervical spine. CMC and ICC values were high with ranges of 0.97-0.98 and 0.87-0.92 respectively (Theobald et al. 2012). CMC is a measure of how well a certain variable can be determined using a linear function of a set of another variables. Similar repeated reliability studies have not been undertaken for other areas of the spine but inertial sensors have been used successfully in lumbar motion and posture (Wong and Wong 2008) and measuring head movement for neck pain assessment (Jasiewicz et al. 2007). Inertial sensors have been validated against both electromagnetic systems (Jasiewicz et al. 2007; Saber-Sheikh et al. 2010) and opto-electronic systems (Wong and Wong 2008) and have been shown to have high correlation. Saber-Sheikh et al (2010) found mean errors between the two systems of -0.69˚, -0.4˚ and -0.28˚ for the x-, y-, and z-axes respectively thus demonstrating the validity of inertial sensors with respect to an electromagnetic system. Jasiewicz et al (2007) also demonstrated the device’s validity with respect to an electromagnetic motion analysis system finding very high values of cross-correlation (0.97-

0.99). There were also very low root mean square errors of 0.7˚ to 2.5˚. This was concluded with respect to cervical spine ROM and thus, inertial sensors are a valid device for the measurement of spinal kinematics.

There are some inherent problems with each component of inertial sensors. Accelerometers can only gather data in two planes simultaneously, gyroscopes suffer from drift, and data from magnetometers is affected by the presence of ferrous metals. The integration of these three components, however, along with corrective mechanisms such as Kalman filters means that three dimensional kinematic data can be obtained as well as the time history of motion. Thus, this device has the potential for being a good technique for the application in question.

In conclusion, a string of inertial sensors would provide a useful method to analyse dynamic spinal motion if spaced minimally apart along the whole length of the spine.

2.1.3. Summary of Kinematic Measurement Devices

There are a number of techniques available that allow spinal kinematics to be measured. Table 1 gives an overview of the different techniques outlined in the review of the literature with the key aspects of kinematic measurement systems being evaluated. Each aspect is given a score out of 10 as a measure of how vital it is for the given application. If a particular feature cannot be scored on a linear scale and is either a 'yes' or 'no' option, the score given was 10 if the device does have that capability and 1 if it did not. For features that are difficult to implement, such as moving an opto-electronic system to a new environment, a score of 5 was given. This is because it is possible to move the system, but it is difficult and time consuming therefore, a score of 10 would not reflect the true nature of the difficulty of the task. For the dimensions column a score of 1 was given to devices that only measured 1 dimension, 5 to those that measured 2 and 10 for those that measured all 3 simultaneously. The scores are then totalled for each device in the right-hand column to determine which device is best suited for the given application. The device chosen was the inertial sensors which is shown in bold italics.

Table 1 – Summary of motion analysis systems to measure spinal kinematics with scores on key features of measurement techniques

Motion Analysis Technique	Portability	Dimensions (1/2/3)	Curvature (Y/N)	Analysis of Multiple Segments (Y/N)	Cost (1 – Expensive; 10 – Cheap)	Line of Sight Not Required? (Y – 1/N – 10)	Static (1)/Dynamic (10)	Total
Inclinometer	10	1	1	1	10	1	1	25
Electro-gonimeter	10	5	1	1	7	1	10	34
Electromagnetic	1	10	10	10	1	10	10	53
Opto-electronic	5	10	10	10	1	1	10	47
Flexicurve	10	1	10	1	7	10	1	40
Inertial Sensors	***10***	***10***	***10***	***10***	***6***	***10***	***10***	***66***

The individual components of inertial sensors have some inherent limitations but when used in conjunction with each other, they overcome the majority of these problems. By assigning a scoring system to various features of kinematic analysis techniques, it allowed all methods to be evaluated and a conclusion could be made as to which method would best suit the desired purpose. With a total score of 66/70, inertial sensors were the best option for use during rugby scrummaging and meets some important criteria, such as being able to measure three-dimensional dynamic activity without the need for line of sight. Therefore, inertial sensors were chosen as the method of kinematic analysis and a set was identified (3AMG, ThetaMetrix, Waterlooville, UK) to be used and validated.

2.2. Review of Force-EMG Relationship Literature

Electromyography (EMG) is a method by which researchers and clinicians can monitor the electrical activity of muscles when they contract. The electrical signals produced by the muscles can be measured and these signals are a function of time and can be described in terms of amplitude and frequency. EMG signals are dependent on a number of anatomical and physiological properties relating to the muscle being monitored. These signals can be measured either on the skin surface using surface electrodes or invasively using needle electrodes but the signal requires a great deal of signal processing. Such signal processing steps include amplification of these very low amplitude signals, rejecting unwanted signals, and filtering. Furthermore, the amplitude of the EMG signal has a relationship to force although the nature of this relationship could be described as contentious as many different researchers report different relationships (Keshner et al. 1989; Kristina Schüldt and Harms-Ringdahl 1988; Swaminathan et al. 2015).

2.2.1. The Force-EMG Relationship

The relationship between EMG and force has been a subject of interest for many years (Lippold 1952). The nature of this relationship is widely debated. Although a qualitative relationship can be seen between the two having processed the EMG signal, a quantitative relationship is much more difficult to establish. The EMG signal is a product of a number of anatomical, physiological and technical factors, which is why a quantitative relationship is so problematic. Technical factors affecting EMG signals are many. Such factors include skin preparation, electrode placement, choice of EMG equipment and smoothing/filtering techniques. Even if these technical factors could be standardised it is still a difficult task to create such a relationship as the tissue between the detection electrode and the muscle of interest is non-homogenous (De Luca 1997). It is particularly difficult to establish a standard procedure for the processing of EMG data. There are two widely used methods which are the root-mean-square (RMS) method and the linear enveloping method. The RMS method is when an envelope of the signal is calculated using a moving ‘window’ of data. An RMS value for each window of data is calculated and then the same is done for the next window and so on. The linear envelope method is a combination of rectifying the data and then low pass filtering it. This latter method is widely used in a variety of studies (Netto and Burnett 2006; Burnett et al. 2008; Swaminathan et al. 2015); however, there are more problems associated with this method as filtering techniques are not standardised. Different authors may use different types of filter, different filter orders and different cut-off frequencies which further adds to the problem of standardising the method of EMG data analysis. The most widely used linear envelope filter is the Butterworth filter. This type of filter is designed so that that it has a flat frequency response in the passband.

The observation that EMG amplitude increases as muscular force increases is indeed correct, but this is a qualitative statement. For example, if the question to be answered is: ‘is a muscle more active during task X when compared to task Y’, it is possible to determine whether this is the case using EMG. If the question, however, is ‘to what extent is a muscle more active during task X when compared to task Y’, this is a much more difficult question to answer, quantitatively, with precision. If the researcher wants to demonstrate a quantitative relationship between EMG and force then the muscle in question should be limited to only isometric contractions. One of the most significant reasons for this is that the change in muscle length during dynamic contractions affects the EMG signal and the force produced. Therefore, its relationship with force is affected. To further complicate the relationship, researchers have found that the linearity of the trend differs between muscles (De Luca 1997). Additional problems include the manner in which soft twitch and fast twitch muscles contract and muscular cross-talk. Muscular cross-talk is when surface electrodes detect muscle activity that is produced by muscles not directly being monitored. That is, as muscular exertion increases, additional muscles are often recruited to aid in force production and these are often not being monitored by the EMG equipment. A specific example of this problem is monitoring deep/surface muscle activity. Surface electrodes may just be monitoring surface musculature or may also be monitoring deep muscle activity and quantifying the magnitude of these contributions is extremely difficult.

One of the most difficult aspects to control when investigating a certain muscle is the recruitment of other motor units to support the contraction. If a motor unit, some distance away from the electrode, is recruited to support force production, the EMG signal will only marginally increase, if at all, compared to the much larger increase in force. On the other hand, however, if a motor unit is recruited close to the electrode, the relative increase in EMG signal will be greater than the relative increase in force. Additionally, as muscle force increases over and above the level of a recently recruited motor unit, the rate at which the motor unit fires will increase but the magnitude of force contribution of that motor unit will not. This will negatively affect the EMG-force relationship. These factors provide a major hurdle when trying to relate EMG to force. The desire to be able to estimate force, to some level of accuracy, however, in situations where direct force measurement is not possible, can be extremely important in determining both the health and tolerance of muscles and joints. Therefore, in spite of the numerous problems when attempting to relate EMG and force, EMG may currently be the best solution to determine muscular force production when direct force measurement is impossible.

Even with the various problems with attempting to relate EMG and force, many researchers have attempted to do so with varying results. The relationship between the two variables has been described as linear (Nigg and Herzog 2007; Keshner et al. 1989; Queisser et al. 1994) and non-linear (De Luca 1997; Sommerich et al. 2000; Solomonow et al. 1990) with Solomonow et al (1990) describing a slight sigmoid curve to their polynomial model. If some relationship can be established between the two, it may prove invaluable in evaluating muscle activity, and therefore force, during in-field activities where direct force measurement is impossible.

2.2.2. Force-EMG Models

Force-EMG models cannot strictly be categorised into two distinct types as previous research has found some regions of the relationship are linear and some where it is not. Nigg and Herzog (2007) showed that the relationship between EMG and force is linear for the range of contractions in the mid-range (~20-75 %MVC). Figure 5 shows this relationship for normalised EMG vs mean force. It can be seen that contractions at the lower and upper range of the muscular contraction exhibited a slight curve as opposed to the linear region in the mid-range. At the low range, the accuracy of the EMG signal may be compromised owing to the small number of motor units firing at any one time thus making it difficult to detect the signal. At the high range, a large number of motor units will be firing and the EMG system may be picking up muscular cross-talk from muscles using to aid the contraction as it approaches a maximum. These are possible reasons for the non-linearity observed at the low and high end of the force-EMG curve in this study. It should be considered, however, that the force-EMG relationship may not be as simple as certain regions being linear and some being non-linear. The relationship shown in Figure 5 may possibly be a third order polynomial relationship and so it is important to consider the dataset as a whole as opposed to isolating specific regions. The authors, however, grouped all the participants data to create this force-EMG relationship and this has been shown to be an inaccurate method of creating such a relationship (Queisser et al. 1994). Thus, the nature of this relationship may not be applicable to all cases.

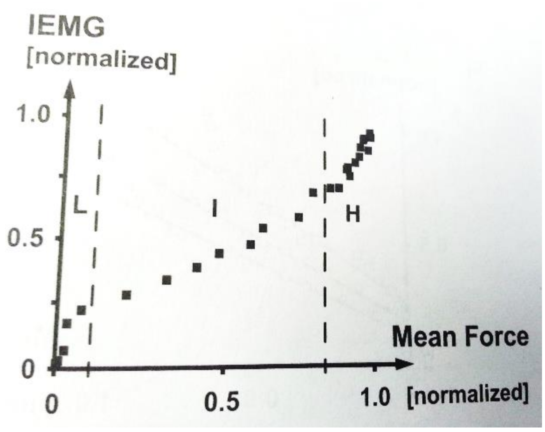

Figure 5 – Force-EMG Relationship (Nigg and Herzog 2007). L – Low exertion; I – Intermediate exertion; H – High exertion. Reproduced from Biomechanics of the Musculoskeletal System by Nigg BM, Herzog W, 2007, Chichester (UK), John Wiley & Sons.

Furthermore, when considering the relationship at different muscle lengths, the slope of the line changed. Thus, it is extremely important to keep the contractions as isometric as possible when trying to develop a force-EMG relationship (De Luca 1997). If contractions are not isometric, then another variable is introduced making it even more difficult to develop a relationship.

Solomonow et al (1990) investigated the force-EMG relationship of the isometric contraction of the gastrocnemius muscle. Similar to Nigg and Herzog (2007), the relationship was found to be linear for a certain level of muscle contractions. The method of least squares ($p<0.05$) was used to fit the best linear regression polynomial to the data. For contractions of up to 50% MVC, the relationship observed was linear. For contractions from this point up to MVC, there was a progressive increase in non-linearity. The reason for this may be muscular cross-talk which was not quantified by the authors and is also extremely difficult to quantify. Therefore, it may be important, if possible, to monitor the muscle activity of surrounding muscle to determine their contribution to force production. It was also observed that increasing the contraction rate did not change the nature of the force-EMG relationship.

Keshner et al (1989) investigated neck muscle activation patterns during isometric head stabilisation. One aspect of the study was looking at the relationship between EMG activation and force. With a linear increase of stimulus force in a given direction, the muscle activity steadily increased in a linear fashion. Figure 6 shows the force-EMG relationship of the semispinalis, splenius and sternocleodiomastoid muscles developed by Keshner et al (1989) for two subjects. The numbers to the right of each curve indicate the orientation of the imposed force. It can be seen from the figure that there was a linear response from the semispinalis (0) and sterocleidomastoid (180) for both subjects. The authors found that a linear increase in force in the preferred and directly opposite to the preferred direction corresponded to a linear increase in EMG activity. When force was applied in a non-preferred direction, there was a non-linear response in muscle activity. This was often characterised by a large increase in muscle activity when increasing force from the penultimate to final weight (6-8lbs). This is because at a higher load, muscles not directly involved in stabilisation at lower loads are required to become active in order to stabilise the head and therefore a non-linear response is observed.

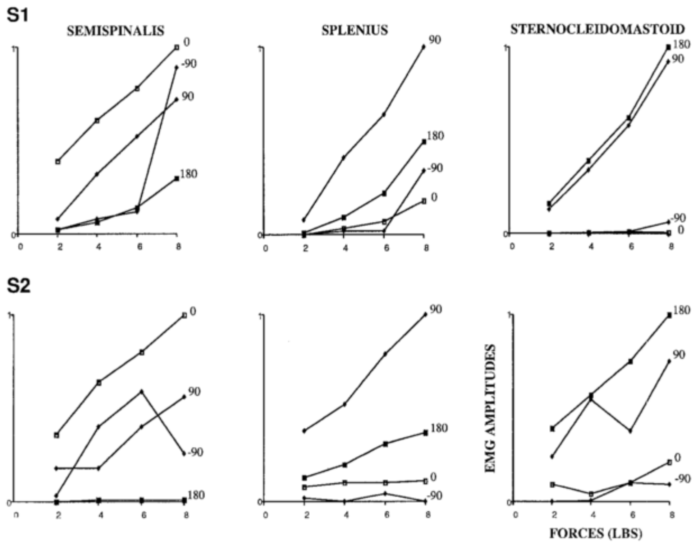

Figure 6 – Force-EMG relationship for different muscles of two different subject, S1 and S2 (Keshner et al. 1989). Reproduced from "Neck muscle activation patterns in humans during isometric head stabilization" by Keshner E, Campbell D, Katz R & Peterson B. 1989. Experimental Brain Research. 75(2), pp. 335-344.

A statistical test for linearity was also performed for each subject which supported the observed linear relationship in the direction of head motion stabilisation ($p<0.05$). First, second and third order polynomials were fitted to the force-EMG curves with the linear response yielding the best relationship in the preferred orientation. The test for linearity was not significant ($p>0.05$) for the non-preferred directions. These findings highlight the fact that when considering the force-EMG relationship, the direction in which the muscles stabilise and the isometric nature of contractions are important factors to consider.

Similar to the previous study, Queisser et al (1994) investigated the relationship between EMG amplitude and isometric extension torques of the neck muscle at different positions of the cervical spine. For each of the four muscles investigated, a coefficient of determination was calculated to quantify the goodness of fit for a linear model. R^2 values ranged from 0.714-0.949 indicating a strong linear relationship between EMG amplitude and isometric neck extension torques. It should be noted, however, that the strongest relationship was for the semispinalis muscle (R^2 = 0.926-0.949) and other muscles did not have such a strong linear relationship. For example, the levator scapulae showed R^2 values ranging from 0.714-0.797. Figure 7 shows the force-EMG relationship developed by the authors for two different participants. For the splenius capitis, the regression line was found to be linear although it can be seen that the gradient is very different for each participant. The study investigated whether or not it is possible to pool individual regression equations as this would be extremely useful for practical applications and save time during data analysis. It can be seen from this figure, however, that pooling the data would incur large prediction errors as each participant demonstrates a different relationship even if they both are linear. Normalisation of the data, however, resulted in reduced between-subject-variation, but it still left relatively large errors when using the grouped regression data. R^2 values ranged from 0.714-0.949 indicating a strong linear relationship between EMG amplitude and isometric neck extension torques. It should be noted, however, that the strongest relationship was for the semispinalis muscle (R^2 = 0.926-0.949) and other muscles did not have such a strong linear relationship. For example, the R^2 values for the levator scapulae ranged from 0.714-0.797. The data was normalised relative to maximum voluntary contraction (MVC). MVC was determined prior to each trial by gradually increasing force production up to maximum. The participants were then given adequate time to rest to avoid the effects of fatigue. Even when EMG and torque were normalised, individual predictions varied by up to 15% of MVC indicating that pooling the data to reduce between-subject variation is not a reliable method.

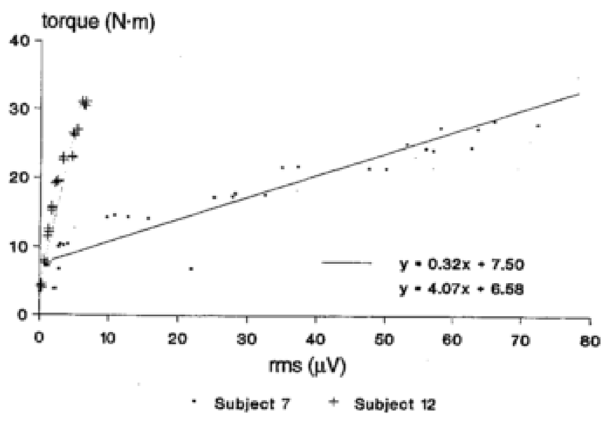

Figure 7 – Force-EMG relationship for two participants for the splenius capitis muscle (Queisser et al. 1994). Reproduced from "The relationship between the electromyogram-amplitude and isometric extension torques of neck muscles at different positions of the

cervical spine" by Queisser F, Ruthner R, Brauer D & Seidel H. 1994. European Journal of Applied Physiology, pp. 92-101.

In contrast to the linear relationships discussed previously, Schüldt and Harms-Ringdahl (1988) found a non-linear relationship between the cervical erector spinae and extension moment when normalising both variables. This may be because the authors were considering extension moment as opposed to force production which will be dependent on the muscle length. Therefore, this may have had an effect on the relationship produced. Figure 8 shows the moment-EMG relationship of all 10 participants. It can be seen that the relationship differs for each participant and the results indicate a non-linear relationship between the two variables.

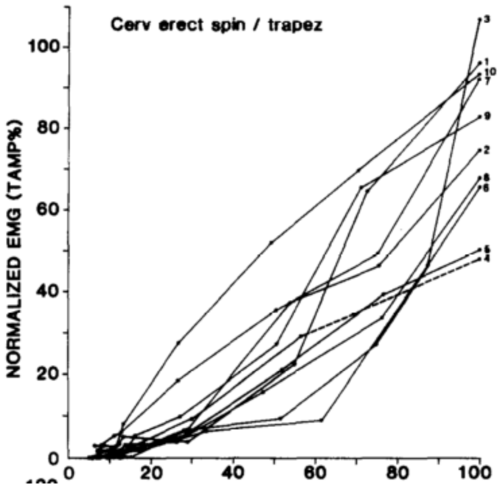

Figure 8 – Moment-EMG relationship of cervical erector spinae for all 10 participants (Schuldt and Harms-Ringdahl 1988). Reproduced from "E.m.g./moment relationships in neck muscles during isometric cervical spine extension" by Schuldt K, Harms-Ringdahl K. 1988. Clinical Biomechanics. 3(2), pp. 58-65.

The non-linear relationship was observed most of the participants, but some participants did indeed show linear relationships between the two variables. For the nonlinear relationships, the authors drew an 'assumed' linear line of best fit on the moment-EMG curve. It was found that for all the muscles investigated, the recorded EMG values were significantly below this assumed line of best fit. The strength of this study is that the authors did not use grouped data to create a relationship.

Relating EMG to force with any degree of accuracy is extremely difficult and the nature of the relationship is always a subject of debate. Taking into consideration the issues raised in previous studies, there were a number of factors that were considered during this thesis. When attempting to relate these two variables, the contractions must be isometric and the joint should not move position causing a change in muscle length as this results in different gradients of the model (Nigg and Herzog 2007; De Luca 1997). Thus, relating these studies to scrummaging is difficult as there is indeed some motion of the neck during scrummaging. A method involving the use of isometric contractions in a controlled environment, however, may be applicable when motion is minimal. The prime motion, the muscle in question results in, must also be considered. If, when the muscle contracts, it results in neck extension, then it is this isometric contraction that should be considered as other stabilising muscles may show a different relationship to the muscle of interest (Keshner et al. 1989). Finally, group results should not be used to predict individual results, since this can introduce large errors (Queisser et al. 1994). As can be seen in the subsequent chapters, individualised force-EMG curves were used as a tool to determine force from EMG data alone.

2.3. Rugby Injuries

This section reviews the literature pertaining to rugby injuries in general with specific regards to injury prevalence, the game events that cause large proportions of injuries and the specific playing positions within a rugby side that are injured.

2.3.1. Injury Prevalence

Before reviewing the injury prevalence literature, it is noteworthy to mention that it is difficult to analyse these types of studies because of the use of various difference units depending on the study in question. Such units used include incidence rate ratio (IRR), injuries/1000 player hours, injuries/player-season, injuries/1000 player match-hours and injuries/100,000 players which makes direct comparison of these studies problematic.

Incidence rate ratio was a measure used to compare the incidence rates of certain game events. Injuries/1000 player hours is defined as the number of injuries that occur for all types of training and match play. That is, if a squad of 30 players took part in 10 hours training and each played 1 hour of a match in one week, the number of player hours would equate to 330 player hours. Injuries/player-seasons is the number of injuries that occur over a playing season. Thus, when 1000 player-seasons are quoted, it could mean that 500 players were observed over two full playing seasons, or 1000 players were observed over one full playing season. Injuries/1000 match hours are calculated in the same manner as player hours. Injuries/match playing hours is slightly different to the previous term as it is only the time played per player that is taken into account.

It would be beneficial to be able to convert each of these statistics to a standard unit to allow for direct comparison but this is not as straightforward as it may first seem. When trying to convert a specific statistic, a number of factors must be taken into consideration, some of which are difficult to quantify. These factors include playing level, intensity of training, type of training, number of training sessions per week, likelihood of injury depending on playing position. Particularly intensity of training and likelihood of injury depending on playing position are very difficult variables to quantify and thus, it was not possible to convert these statistics to one standard unit.

Table 2 summarises the data from a number of different rugby injury prevalence studies. The data indicates a wide range of injury prevalence which could be dependent on a number of factors. One such factor which is mentioned in a number of different studies is playing level. It is suggested in these studies that the higher the playing level, the greater the prevalence of injury (Doyle and George 2003; Lee and Garraway 1996; Roberts et al. 2013). With regards to the statistics presented in Table 2, they were averaged for the number of seasons for which data was collected. For example, Haseler et al (2010) and Nicol et al (2011) collected data over one playing season whereas Palmer-Green et al (2013) collected data over 2 seasons. Regardless of the study, it is evident that rugby has a relatively high injury prevalence rate at all levels of the game, from youth to professional.

Table 2 - Injury prevalence studies with cohort investigated, statistic provided, common injury sites, game event and factors affecting injury

Authors	Cohort	Statistic	Units	Factors affecting Injury
Doyle and George (2003)	England Women's Rugby	27	Injuries across 35 players in a squad	Age, weight, playing position
Peck et al (2013)	Men and Women's Amateur	1.30 (95%CI – 1.09-1.54)	Incidence rate ratio (IRR)	Gender
Haseler et al (2010) Nicol et al (2011) Bleakley et al (2011) Palmer-Green et al (2013)	Youth Rugby	24/1000 (95% CI – 16.4-31.3) 10.8/1000 27.5-129.8/1000 35/1000	Injuries/Player hours Injuries/Player hours Injuries/Match hours Injuries/Player hours	Age (aggression, competitiveness), playing level
Lee and Garraway (1996)	School Senior Amateur	School: 86.8/1000 (95% CI – 73.4-100.2) Senior: 367.0/1000 (95% CI – 339.4-394.6)	Injuries/Player-seasons	Playing level
Bottini et al (2000)	Senior Amateur	2.4	Injuries/weekend	N/A
Bird et al (1998)	Senior Amateur	10.9/100	Injuries/Player games	N/A
Clark et al (1990)	Senior Amateur	1/60	Injuries/Match playing hours	N/A
Roberts et al (2013)	Semi-professional Amateur Recreational	16.9/1000 (95%CI – 19.7-23.6) 16.6/1000 (95% CI – 15.2-17.9) 14.2/1000 (95%CI – 15.2-17.9)	Injuries/Player match-hours	Playing level
Bathgate et al (2002) Williams et al (2013) Targett (1998) Brooks et al (2005b) Schwellnus et al (2014)	Professional	69/1000 81/1000 120/1000 91/1000 9.2/1000	Injuries/Player hours	N/A

N/A – this information was not provided for the specific study

2.3.1.1. Scrummaging

With rugby being such a physical contact sport, there are number of game events that can result in injury. The most common game situations that result in injury include tackles, rucks, mauls, collision events and scrums. This thesis, however, is focussed on scrummaging and therefore only scrummaging injury literature is reviewed.

The scrum has been shown to be the event that carries the highest risk of injury per event. Per 1000 events, 213.2 days of training/match play were lost owing to scrummaging injuries. Scrums occur much less frequently than tackles but carried a 60% greater risk per event (Fuller, Brooks, Cancea, et al. 2007).

Taylor et al (2014) found that 31% of scrums in a season of English professional rugby union collapsed. Collapsed scrums had a significantly greater risk ($p=0.04$) of causing injury than scrums that did not collapse. The reason collapsed scrums are so high risk is that the players of the front row have very little control over the way in which they collide with the turf. The hooker, in particular, is bound into position and thus, aside from quickly repositioning their head as the scrum collapses, have very little control over their fall. Therefore, this event provides a significant risk of hyperflexion injuries with or without rotation/compression (Scher 1982) as players are already in a position of cervical flexion during normal scrummaging and may be forced into hyperflexion as they are driven into the turf. The scrum is a high risk contact event and consistently causes injuries but is not the event that causes the greatest number which is most likely owing to the fact that much fewer scrums occur during a game than tackles or rucks. In women's rugby 8.9% of injuries were from the scrum (Taylor et al. 2011), in schoolboy rugby it was 11.9% (Sparks 1985), and 11% in professional men's teams (Brooks, Fuller, Kemp, et al. 2005a).

When considering spinal injuries specifically, however, the scrum causes a much more significant proportion of injuries with 36% being found in U19 rugby in the British Isles (Maclean and Hutchison 2012). It should be noted, however, that this particular statistic is from players who were admitted to a spinal trauma injury clinic. Thus, it is a large proportion of severe rugby spinal cord injuries that result from the scrum. Similarly, Hendricks et al (2014) found that in South African rugby over a 4-year period, the tackle accounted for 47%

of overall injuries and the scrum 33%. Even though the tackle resulted in a greater proportion of overall injuries, it was the scrum that had a greater proportion of permanent injury outcomes with 70% (n=20) of scrum-related injuries resulting in some form of catastrophic spinal cord injury; the tackle resulted in 57% (n=28). These proportions were calculated from the number of injuries resulting from each game situation. Furthermore, of the scrum-related injuries, 85% occurred during matches and 95% were at the amateur level indicating that competition and playing level may have an effect on scrum injury risk. This data was collected through the BokSmart programme where it was a requirement to report all injuries to the governing body, the South African Rugby Union.

Although numerous studies show that the scrum is a phase of play that results in a substantial proportion of injuries, other research suggests that it only accounts for a very minor proportion of injuries. Bathgate et al. (2002) found that in elite Australian rugby the scrum only caused 2% of injuries and Holtzhausen et al. (2006) found that in South African Super 12 teams, the scrum only accounted for a small number of minor injuries. Scottish schoolboy and women’s rugby have also reported low incidence of scrummaging injuries with only 5.4% (Nicol et al. 2011) and 5.6% (Taylor et al. 2011) of injuries respectively. Therefore, there is some conflicting evidence in the literature as to the proportion of injuries that result from the scrum. It is a phase of play, however, that can result in catastrophic spinal cord injury and this is widely acknowledged in the literature.

An idea that was recently raised was that injury epidemiology studies may be underestimating the neck injury incidence resulting from the scrum (Brown et al. 2014). There are two main methods of calculating ‘exposure’; the ‘athlete at risk’ method and the ‘athlete participation’ method. When considering the potential outcomes depending on the various possible input variables both these approaches can produce a number of different results. For example, if you use the ‘athlete at risk’ method and consider all 15 players the authors found that the scrum injury incidence rate was 10.8/1000 player-hours (95% CI 8.6-13.1). Since this includes back players, who are not part of the scrum, this offsets the outcome to a reduced risk. So if you consider just the 8 players of the scrum the injury incidence rate was 20.3/1000 player-hours (95% CI 16.1-24.5). This, however, does not account for the fact that the front row are by far at most risk and using just these three players results in an injury incidence rate of 54.2/1000 player-hours (95% CI 43.0-65.3). This method relies on which players the researcher considers to be 'at risk' from the scrum. Yet another value is yielded if the 'athlete participation' method is used. The authors suggest that researchers must consider which players are at risk before calculating injury incidence as this affects the outcome. Despite injury rates not being significantly different when using the different methods, the method used is important as injury incidence may be underestimated by all the methods (Brown et al. 2014).

Despite some conflicting reports in the literature about the injury incidence of the scrum, the pertinent point is that the scrum is a high risk event and is more likely to cause an injury, potentially severe, compared with other match events.

2.3.1.2. Playing Positions Injured

Table 3 shows injury prevalence data according to specific playing positions that were found to suffer from the greatest proportion of injuries in rugby sides. It can be seen that in every study, it is always a forward playing position that is found to be most injured. The specific forward playing position varies from study to study but it is always from the forwards. Furthermore, it can be seen that in 3 out of the 6 studies the front row were found to suffer from the greatest number of injuries.

Table 3 – Injury prevalence of specific playing positions in a variety of studies

Authors	Cohort	Playing Position Most Injured	Injury Statistic
Sparks (1985)	School	Front Row	>50%
Nicol et al (2011)	School	Front Row	53.3%
Best et al (2005)	Professional	Props	17%
Jakoet and Noakes (1998)	Professional	Hooker Second Row	7% 25%
Clark et al (1990)	Male Adult Amateur	Props	6%
Targett (1998)	Professional	No. 8	N/A

N/A – injury statistic not provided

2.3.2. Scrummaging Injuries

This section focuses solely on scrummaging injuries, how prevalent they are, spinal injuries resulting from the scrum, possible injury mechanisms and injury prevention strategies.

Recently, Trewartha et al (2015) reviewed the scrummaging literature relating to both injuries and biomechanics. This study concluded that, at most, 10% of all rugby-related injuries are caused by the scrum and the majority of this 10% are injuries that are characterised as moderate in severity. Although only a small proportion of all rugby injuries result from the scrum, the largest proportion of spinal cord injuries in rugby (40%) result from this game event. Furthermore, there is evidence that suggests scrummaging to be a contributing factor to chronic degeneration of the cervical spine but there is a lack of longitudinal data to say this definitively. Thus, it is evident that rugby scrummaging is likely to be a contributing factor to chronic spinal injuries and also is the cause of a large number of acute injuries. In this context, acute injuries are those that occur suddenly and are likely to be the result of a collapsed scrum or improper scrum engagement. Chronic injuries, however, are those that do not result in any significant trauma in the short term, but, over a period of time, the repeated microtrauma players are exposed to cause significant damage. In engineering terms, it is the difference between instantaneous loading (acute) and cyclical/fatigue loading (chronic) of a material until some failure occurs.

2.3.2.1. Injury Prevalence

The scrum is an event that poses a high risk of injury, albeit relatively low incidence (Fuller, Brooks, Cancea, et al. 2007; Roberts et al. 2013; Brown et al. 2014), particularly to front row players. It has been shown that the overwhelming majority of injuries (88-91%) from the scrum occurred to the players of the front row (Brooks, Fuller, Kemp, et al. 2005a; Taylor et al. 2014). The type of injury sustained from scrummaging varies greatly with the cervical spine, lumbar spine, shoulder musculature, and calf musculature all being frequently cited as common injury sites. It is the cervical spine, however, that receives the greatest attention in the literature owing to the catastrophic nature of such injuries. Despite the wide coverage given to cervical spine injuries in the literature, they are not the most common. Calf muscle injuries are the most common scrummaging injury and also result in the greatest number of days absence from training and match play. The next most common injury site is the shoulder which caused 33% of absence and then cervical spine injuries at 15% (Brooks, Fuller, Kemp, et al. 2005b; Brooks, Fuller, Kemp, et al. 2005a).

With regards to cervical spine injury during the rugby scrum, there are two types; acute and chronic. Acute injuries are those that are most likely to happen from two different scrummaging scenarios; the collapse or improper engagement. From 1970-1996, significantly more cervical spine injuries occurred during engagement (64%) than collapse (36%) ($p<0.002$) in American rugby (Wetzler et al. 1998). Contrary to this however, Du Toit et al (1999) stated that 90% of scrummaging injuries were from scrum collapse. It was also observed that 50% of all cervical spine injuries in rugby were from some event relating to the scrum indicating the high risk nature of this event. These two situations, improper engagement and collapse, primarily affect members of the front row. In America, 78% of cervical spine scrummaging injuries were to the hooker and of the injuries that occurred during scrum engagement the hooker was injured significantly more than either prop ($p<0.003$) (Wetzler et al. 1998).

When considering U19 players from the British Isles, it was found that 92% of injuries that occurred from the scrum resulted in some form of spinal cord injury with 61% of these completely severing the spinal cord. Spinal cord injury was also found to be significantly more common ($p<0.001$) as a result of the scrum rather than the tackle (Maclean and Hutchison 2012). This particular statistic, however, should be given in context as it was a spinal trauma treatment centre in which the data was collected.

In professional English rugby, 31% of scrums in competitive matches collapsed. It is the scrum collapse that is considered to be one of the most injurious events. Furthermore, the injury incidence associated with the collapsed scrum (8.6/1000 events) was significantly higher ($p=0.04$) than the injury incidence where collapse did not occur (4.1/1000 events). Of all the injuries observed that were scrum-related, 41 in total, an overwhelming 88% of injuries occurred to the front row; 16 to the tight-head prop, 11 to the loose-head prop, 3 to props who did not specify position and 6 to hookers. This particular study goes against the view that the hooker is the most injured player in the scrum but it does highlight the high risk of injury to front row players from scrummaging (Taylor et al. 2014).

The epidemiological studies in the literature show the scrum to be an event that results in a significant number of injuries to different areas of the body, the most concerning of which is the cervical spine. Furthermore the hooker suffers from a large number of injuries from the scrum and is therefore the player of most interest in this thesis.

2.3.2.2. Spinal Injuries

Cervical Spine Injuries

Scrummaging causes 35-50% of all cervical spine injuries in rugby union (Bohu et al. 2008; Dunn and van der Spuy 2010; Maclean and Hutchison 2012; Wetzler et al. 1996; Secin et al. 1999). Of these injuries, the front row players represent up to 83% of cervical spine injuries in the scrum (Bohu et al. 2008; Brooks, Fuller and Kemp 2005; Brown et al. 2013) with those in the hooker position most susceptible (Bohu et al. 2008; Secin et al. 1999; Wetzler et al. 1996; Wetzler et al. 1998; Brown et al. 2013). Moreover, Secin et al (1999) reported that hookers are at a statistically greater risk ($p<0.01$) of sustaining a disabling injury of the cervical spine, potentially as a consequence of the most commonly used method of binding, the over-bind. Subsequently, in instances of the scrum collapsing, the hooker is powerless to control their impact with the ground, aside from repositioning their head, which often gets forcefully driven into the turf. This problem may be the reason the hooker is at such a high risk of sustaining a disabling injury of the cervical spine. Anecdotally at least, this presents a catastrophic injury risk (Scher 1982; Scher 1991; Scher 1990a).

It has been suggested that lateral and vertical forces during the scrum cause premature degeneration of the cervical spine as well as acute injuries (Scher 1990b; Milburn 1993; Milburn 1990). Triantafillou et al (2012) raised some pertinent points evaluating the cause of premature cervical spine degeneration. They note that in collision sports, it has been repeatedly observed that players who sustain the greatest frequency of repetitive loading have higher rates of cervical spine degeneration although there is no statistical evidence to support this. The authors suggest a correlation between the two but this does not necessarily mean that one causes the other owing to the lack of statistical significance.

Additionally, the authors question whether the presence of degenerative disease of the cervical spine is a contributing factor to further acute cervical injury if the athlete continues to participate in their chosen sport. This latter point is extremely important when considering the physical nature of rugby, particularly events such as the scrum, where collision cannot be avoided. This sort of disc degeneration that is evident in the front row causes loss of overall ROM of the cervical spine. Other forms of degeneration include the formation of osteophytes (Broughton 1993; O'Brien 1996; Berge et al. 1999). Despite reduced ROM in the three anatomical planes, it has been shown that translational motion of the vertebrae increases with mild levels of degeneration but ROM decreases again when degeneration is severe (Miyazaki et al. 2008). This is particularly pertinent to front row rugby players who suffer from cervical spine degeneration (Broughton 1993; Scher 1990b; O'Brien 1996; Berge et al. 1999; Castinel et al. 2010; Hogan et al. 2010) as it is likely that their cervical spine has decreased stability for this type of translational motion.

A number of case studies have been presented focussing on cervical spine injuries in rugby players. One such case involved an inexperienced player playing in the front row who sustained an anterior dislocation of C5/C6 with bilateral locking of facets which occurred when the scrum collapsed (Scher 1982). This injury occurred as a result of forced hyperflexion and this particular injury resulted in complete paralysis below the level of injury. Inexperienced players playing the front row have since been stopped by World Rugby as players now require an adequate level of training to play in this position (IRB 2013). Another similar case was caused on engagement where the hooker's head impacted with the opponent's shoulder and was forcibly flexed and rotated. This injury also resulted in complete paralysis below the level of injury (Scher 1982). Similarly, a prop forward was injured on scrum collapse which resulted in forced hyperflexion of the neck and caused complete paralysis below the level of fracture. On examination post injury, the player was found to have congenital vertebral fusion of the vertebral bodies and posterior aspects of C2 and C3 as well as anterior dislocation at the level of C6/C7 (Scher 1990a).

With regards to chronic injuries, degeneration of the cervical spine is likely to result in decreased ROM. Cervical ROM over a playing season was found to decrease which was much more significant in forwards than backs (p=0.03). This increased effect was suggested to be a cause of forwards taking part in more contact events such as scrummaging. The ROM exhibited by forwards at the end of the playing season is comparable to the elderly or patients with acute whiplash. More specifically, rotation did not decrease significantly (p>0.05) but right lateral bending did (p=0.04) (Lark and McCarthy 2010) although it is unclear why only right lateral bending decreased and left lateral bending did not. The same authors have shown that cervical ROM decreases over the course of a single game (Lark and McCarthy 2009) but this is most likely owing to muscle soreness as opposed to potential vertebral microtrauma. Microtrauma, however, may be a cause in the study over the course of a playing season. A similar study supporting these findings shows that rugby players, particularly forwards, demonstrate increased prevalence of neck pain, decreased cervical ROM and reduced head positioning accuracy. Forwards had a greater reduction in cervical ROM than backs and this was attributed to increased involvement in contact events. As well as this, forwards were found to be 7 times more likely to have neck pain than backs and 33% of forwards demonstrated a decreased head repositioning accuracy. All these problems were attributed to increased involvement in contact events (Gemmell and Dunford 2007). Lark and McCarthy (2010) also investigated cervical ROM recovery during the rugby off-season. It was found that both forwards and backs do have an increased ROM given time to recover during the off-season but there was a limited amount of recovery. The authors suggest there is a possibility that the off-season is not long enough to allow significant recovery in neck ROM or, alternatively, that players are not undergoing sufficient or appropriate rehabilitation of the neck to aid an increase in neck ROM.

Neck strength may be a factor in helping prevent cervical spine injury and there has been some evidence to suggest that increasing isometric neck strength reduces this risk. It is important to consider the potential mismatch in playing ability when considering these types of injuries. Hamilton et al (2014) found that adult front row forwards were significantly heavier and had substantially greater isometric neck strength (p<0.001) than their under 18 counterparts. Moreover, in a multivariate modelling analysis of cervical strength, the player's weight (r=0.4) and number of years playing experience (r=0.5) were the only relevant factors in predicting neck strength. This study highlights the necessity to not mismatch players when youth players are playing in the adult game. There has also been some research comparing anatomical dimensions of the neck of the front and second rows to controls. Given the large stature and musculature of rugby players, it is not unreasonable to expect that rugby players have significantly larger necks, particularly those that play in the front row. This, however, was not the case as no significant differences ($p>0.05$) were found for any of the measurements. The authors did however suggest that the small sample size used resulted in the non-significant differences and a larger size would yield more significant results (Crombie et al. 2012).

From the literature it can be seen that a variety of spinal injuries, both chronic and acute, are caused from scrummaging. It is therefore an important problem that warrants further investigation from a biomechanical perspective in order to gain a better understanding of potential injury mechanisms.

Other Spinal Injuries

Although cervical spine injuries are the most common from scrummaging there have been cases of lumbar spine injury. There is evidence to suggest that certain players suffer from radiological abnormalities which may be caused by scrummaging. One case report was a second row player where the lumbar spine was hyperflexed on engagement with the other pack. The player presented with lower back pain and, on diagnosis, it was found that the discs of L4/L5 and L5/S1 were degenerated. An MRI scan confirmed this finding and also suggested a stress fracture of the pedicles of L4 and L5 but no injury to the spinous process. The authors do suggest that it is an unusual injury to occur, however (Jones et al. 2005).

Other authors suggest that front row forwards in particular are at a high risk of lumbar spine (L-spine), as well as cervical spine, injuries (Fuller, Brooks and Kemp 2007). Iwamoto et al (2005) theorised that the dynamic loading patterns in rugby union produce relatively high stresses in the L-spine, particularly on the disc, facet joints and pars interarticularis. This hypothesis was supported by 243/327 (74.3%) of high school front row players having at least one radiological abnormality when investigating low back pain. For front row players during scrummaging, there is an enforced loss of lordosis owing to flexion at the hips. The resulting compression from the player's muscle activity, second row players and opposing front row players all provide a high risk of disc failure (Adams and Hutton 1982; Adams and Hutton 1985a). These studies use cadaveric models to demonstrate that, when flexed, compression is likely to result in disc failure if a threshold is reached which has also been shown by other authors (Aultman et al. 2005). The key point is that flexion and compression above a certain threshold is likely to injure the disc. Further evidence for the vulnerability of the disc is provided by Fuller, Brooks and Kemp (2007) who identified that the most common complaint was lumbar disc injury during scrummaging and more specifically, scrum collapse. In this study, as with others focussing on the cervical spine, front row players were identified as being at the greatest risk of suffering from L-spine injuries. Although L-spine injuries do not have the same potential severity as cervical spine injuries, it is still a problem prevalent in front row players.

2.3.2.3. Injury Mechanisms

This section provides information relating to our knowledge of in vitro injury mechanisms and how these may be relevant to scrummaging injuries. In vitro testing has contributed significantly to our understanding of failure mechanisms in both the lumbar and, more recently, the cervical spine. Resistance to (hyper)flexion has been quantified as being distributed across the spinous processes and associated ligaments, the zygapophyseal joint and the intervertebral disc (ratio: approximately 50/25/25%) (Przybyla et al. 2007). If the scrum collapses and the cervical spine is forced into (hyper)flexion, these structures are likely to be injured if the maximum load that they can support as a construct is exceeded. Resistance to extension has been attributed to approximately half from the zygapophyseal joint, a quarter from the intervertebral disc and the final quarter from the spinous processes. Scrums ‘popping out’ or pre-emptive engagement resulting in (hyper)extension could result in injury to these structures as resistance is provided by them in cases of hyperextension.

The lumbar spine follows a similar picture. In full flexion, resistance is evident from the capsular ligaments of the zygapophyseal joints (~40%), the disc (~30%), ~20% from the supra and interspinous ligaments and ~10% from the ligamentum flavum (Adams et al. 1980). Damage in rapid flexion is seen first in the interspinous and supraspinous ligaments (Adams et al. 1980) followed by the capsular ligaments especially with the addition of lateral bending (Neumann et al. 1992). Eventually the outer annulus of the disc can cause bony avulsion at very high flexion angles (Adams and Hutton 1982).

These experiments, however, only apply a modest amount of compression and only when studying the resistance to flexion/extension. During a scrum collapse front row players are likely to be experiencing high muscle activity in order to prevent collapse and such high muscle activity will increase the level of compressive loading. High compressive loads in the absence of flexion or extension subjects the vertebral endplate to a compressive stress resulting in a greater failure rate at the endplates in the cervical (Przybyla et al. 2007) and lumbar (Adams and Hutton 1982; Brinckmann et al. 1988) regions. Furthermore, the tolerance of the spine to compressive load is dependent on the degree of flexion in the motion segment. Compared with loading in a neutral position, the motion segment flexed to 45° was 50 times more likely to fail under the same compressive load (Gallagher et al. 2006) suggesting the motion segment is weaker in resisting compression once flexed. Therefore, those who take part in sport where copious compression on flexed motion segments is frequent, such as scrummaging in rugby, are at greater risk of failure in response to compressive loading. Moreover, compression applied to the flexed segment is more likely to cause disc prolapse (Adams and Hutton 1982). Whilst it is not known whether similar flexion/compression stresses give rise to similar effects in the cervical spine, it appears likely that owing to the same mechanical responses seen to other stresses, similarity could be determined for the effects on the cervical spine.

Similar to studies of acute injuries, there have been studies examining the fatigue characteristics of the lumbar region. Brinckmann et al (1988) cyclically loaded the lumbar spine in compression and clearly demonstrated that as loading decreased, cycles to failure increased. Also, as the relative loading increased, the probability of finding a fatigue fracture increased for a given number of load cycles. Cyclical loading of the lumbar spine resulted in some form of damage to the vertebral endplate in the majority of cases. Adams et al (2000) also found damage to the vertebral endplate during fatigue testing of the lumbar spine using a complex mechanical loading regime. The result of endplate damage caused reduced pressure in the nucleus pulposis and increased stress concentrations in the annulus, especially in lordotic postures. Endplate damage led to progressive structural changes to the neighbouring intervertebral discs during fatigue testing.

Some studies have analysed the cervical spine of rugby players using MRI and X-ray imaging techniques. MRI images showed players with evidence of advanced osteosclerosis, degeneration of the vertebral endplate, osteophyte formation (Berge et al. 1999), a reduced spinal canal diameter, and degenerative lesions (Castinel et al. 2010). X-ray images have similarly shown evidence of osteophyte formation (Broughton 1993; O'Brien 1996) and also, more generally, widespread degenerative changes of the cervical spine (Hogan et al. 2010; O'Brien 1996). Given the evidence presented relating to the fatigue failure of the lumbar spine, it may be reasonable to hypothesise that the degenerative changes of the cervical spine in rugby players have been caused by similar, repeated compressive loading.

There are a number of mechanisms of acute injury to the cervical spine that can occur during scrummaging as outlined in Table 4. Hyperflexion with or without rotation (Scher 1982; Scher 1990a) and hyperextension (Secin et al. 1999) injuries may result from the scrum although much of the evidence presented for this is in case studies. Almost all the cases presented by Secin et al (1999) show that it is the C4, C5 and C6 spinal levels that are injured whether on engagement or collapse.

Table 4 - Cause, Mechanisms and Description of Injuries during Scrummaging

Cause of Injury	Mechanism of Injury	Description
Charging in/popping out	Hyperextension	When one team pre-empts the engage call causing a disrupted engagement/the result of the loose-head prop pushing inwards towards the opposing tight-head prop causing his chest to 'pop' up and out of the scrum
Collapse	Hyperflexion and/or rotation and/or compression	The scrum crumples resulting in the heads of the front row being driven into the turf.
Wheeling	Combination of lateral bending and rotation cause excessive horizontal shear	2 types: i) excessive pushing on one side of the scrum so that it turns; this is legal. ii) Front rows pulling the scrum to rotate it; this is illegal.
Inefficient technique	Lateral force (suggested cause of premature C-spine degeneration (Scher, 1983))	Poor transmission of forward force causing excessive loading in directions that are not helpful to efficient scrummaging technique

Scrum collapse results in hyperflexion injuries which can occur in isolation or with compression and/or rotation (Scher 1982; Scher 1990a; Scher 1998). In this case, the osteoligamentous cervical spine has been shown to be relatively weak in resisting flexion with the contributions to resistance against forced flexion coming from the interspinous ligaments, apophyseal joints, spinous processes and the cervical disc (Przybyla et al. 2007). The commonality of this injury may be owing to the cervical spine being weakest in flexion. This sort of injury can result in fracture dislocation of the C4/5 or C5/6 junction (Quarrie et al. 2002), unilateral facet dislocation (Dennison et al. 2012; Scher 1998) or facet locking (Scher 1982) in front row players.

In extension, the osteoligamentous cervical spine has been shown to be roughly 25% stronger than in flexion when subjected to compressive load. In hyperextension injuries, most resistance to this forced motion is provided, initially, by the apophyseal joints and then the cervical disc. Therefore these structures are vulnerable during cervical spine hyperextension. Resistance to direct axial compression is provided predominantly from the vertebral endplate, disc and vertebral body (Przybyla et al. 2007). These structures are effectively in series and therefore the loads will be borne through all of them.

In vitro studies testing the failure mechanics of the cervical spine have determined the types of injury sustained during torsional, bending and compressive loading. During axial compression, it was established that damage to the vertebral endplate always occurred. Positioning the spine in flexion, extension or rotation significantly reduced the failure limits of the cervical spine during compressive loading (Aultman et al. 2004). This finding may be particularly important in rugby scrummaging as the neck is likely to be forced into complex positions. Such complex loading patterns would result in a reduced tissue tolerance to compressive loading which may contribute to spinal failure or the development of degenerative changes. A study of dynamic cervical spine injuries confirm that vertebral body damage or fracture is the most likely outcome during compressive loading (Yoganandan et al. 1991). Berge et al (1999) showed that degeneration of the vertebral endplate was present in front row players suggesting that axial compression may be present during scrummaging. Furthermore, posture is a key factor in the initiation of injury with the level of lordosis increasing the likelihood of injury (Holsgrove et al. 2015). Therefore, spinal posture during rugby scrummaging is extremely important as improper posture is likely to increase the likelihood of spinal injury.

There has also been research into chronic injury mechanisms which can be related to the degenerative changes of the spine observed in front row rugby players. Chronic injuries of the cervical spine may result from repeated compression caused by paraspinal muscle activity. This activity results in the loss of intervertebral disc height and resultant increased load bearing on the neural arch and uncovertebral joints (Skrzypiec et al. 2007). With time, this may lead to the development of osteophytes (Kumaresan et al. 2001). These findings are based on cadaveric or modelling studies, however in vivo studies using MRI scans of rugby players have similar findings (Berge et al. 1999; Castinel et al. 2010).

A relatively high incidence of disc narrowing of 35% (Castinel et al. 2010) and 71% (Berge et al. 1999) was observed in rugby players particularly those of the front row. Observation of osteophyte development (83%), apophyseal joint degeneration (74%) and degeneration of endplates (77%) in front row players (Berge et al. 1999) all suggest that these players are exposed to high compressive forces over a prolonged period of time. Hogan et al (2010) similarly found apophyseal joint degeneration in front row rugby players compared to controls with a significant difference at the C2/3, C5/6 and C6/7 levels. A study of two individual cases of cervical degeneration in front row players supports these findings (O'Brien 1996). Anterior and posterior osteophyte formation was observed as well as disc narrowing at the C5/6 level and/or C4/5 or C6/7 level. Broughton (1993) also found the formation of osteophytes at the C5/6 and C6/7 levels of a hooker. It is these types of injuries that are found in the rugby union front row and are most likely a result of cumulative compressive forces over time associated with repeated scrummaging and paraspinal muscle activity.

Currently no methods exist to measure cervical spine loading during live scrummaging that is likely to contribute to chronic changes of the cervical spine. Beginning to quantify these loads may provide an insight into the reasons why front row players, particularly the hooker, suffer from chronic degeneration of the cervical spine.

2.3.2.4. Injury Prevention Strategies

A number of different injury prevention strategies have been implemented over the years to reduce the likelihood of spinal injury particularly with regards to the scrum. These strategies include injury prevention programmes, law variations regarding the scrum and strengthening of the neck musculature.

Scher (1981) previously suggested that strength and conditioning of the neck muscles will reduce the risk of cervical injury. More recently, there have been studies investigating the effect of a neck strength training programme. Naish et al (2013) implemented a specific neck strengthening programme to try and decrease the prevalence of cervical spine injuries in professional male rugby players. When observing the total number of cervical spine injuries and the time lost owing to these injuries, no significant differences ($p>0.05$) were found. There was, however, a significant reduction ($p=0.03$) in the number of cervical injuries during match play. In the 2007-08 season, 11 injuries were observed and in the 2008-09 season only 2 injuries were found. Despite the significant decrease in cervical injuries the authors concluded that it cannot be said with certainty that the neck strengthening programme was the cause of the injury reduction.

Other studies have also investigated the neck strength of rugby players. In extension, forwards have demonstrated a much greater strength than backs (Geary et al. 2013; Olivier and Du Toit 2008) and front row forwards have been shown to have the greatest peak extension torque (Olivier and Du Toit 2008). It has been suggested that scrum stability is achieved by the front row through, in part, neck extension to keep the head stable and also the angle of force application in the vertical plane. This may be a reason that front row forwards performed much better in neck extension strength measurements than other groups. Although previous evidence does not conclusively suggest that increasing neck strength decreases the likelihood of injury, there has been some research to show that this is the case for other muscle groups and other sports. Therefore, a similar approach is suggested for rugby players to try and minimise neck injuries during contact events (Geary et al. 2014). The authors not only suggest neck strengthening but also a regular screening process to allow for early identification of any strength deficits which may be an indicator of the onset of injury.

Another approach that has been adopted by some countries are specific strategies to make the rugby population more aware of certain risk. These include the RugbySmart programme in New Zealand (Gianotti et al. 2009), the BokSmart programme in South Africa (Dunn 2009) and the Mayday procedure in Australia (Poulos and Donaldson 2012). All these programmes were found to be somewhat effective in reducing the risk of spinal injury or, at least, managing the risk of possible further injury in the case of the Mayday procedure.

Increasingly, more and more strategies are being adopted to try and reduce the risk of scrum-related injuries, particularly catastrophic injuries. The results of these studies do appear to have a positive effect as desired.

2.3.3. Summary of Rugby Injury Literature

It is evident from the literature that rugby is a sport where there is a high prevalence of injury owing to its physical nature. Certain game events, such as the tackle and the scrum, have been shown to cause a large proportion of injuries during the scrum. The scrum is regarded as a high risk event and carries the risk of spinal injury particularly if the scrum collapses. Furthermore, a number of cases have provided evidence of such acute injuries to front row players. Not only are acute injuries a risk, but chronic injuries of the cervical spine are also becoming an increasing problem. It is likely that the repetitive loading experienced by front row players during scrummaging contributes to the onset of premature cervical spine degeneration. Therefore, trying to establish potential causes and mechanisms of spinal injury is an important step towards a better understanding of these injuries and, therefore, reducing their occurrence.

2.4. Scrummaging Biomechanics

This section seeks to summarise and critique the current literature available pertaining to the current research on rugby scrummaging biomechanics.

Whilst a rugby scrum aims to quickly, fairly and safely restart the game, each event provides an opportunity for the opposing team to regain possession of the ball by driving forward and thus displacing opposing players. Hence, the generation of a high forward force is a vital component of a successful rugby team. Large forces generated in all three planes are likely to be a major contributing factor to scrum-based injuries especially considering our knowledge of in vitro testing and, as such, this section will consider the force exerted by the whole pack broken down into: i) evaluation of previous scrummaging research, ii) forward force, iii) lateral and vertical force, iv) force distribution across the front row, v) scrummaging kinematics and vi) electromyography during scrummaging.

2.4.1. Evaluation of Previous Scrummaging Research

As previously mentioned, the hooker is one of the most injured positions in a rugby union side and is particularly susceptible to injury during the scrum (Fuller, Brooks, Cancea, et al. 2007; Secin et al. 1999). Previously, research has focussed on quantifying force production during machine scrummaging (Milburn 1990; Milburn 1993; Preatoni, Stokes, et al. 2012) with only a limited amount of kinematic analysis provided. More recently, however, force production has been investigated during live scrummaging, as well as some kinematic analysis (Cazzola et al. 2015). With regards to the machine scrummaging studies, force was determined by instrumenting the shoulder pads of the machine with a force platform, load cell or strain gauge. Strain gauges are a component part of both force platforms and load cells thus making these methods very similar but there are some differences. The main difference is that load cells are used where the force is applied over a small area whereas force platforms are used when the force is to be applied over a wider area. With both these methods, however, the applied load is converted into some form of electronic signal which, depending on the circuitry used, could be a voltage, current or frequency change. Although these are all reliable and widely used methods in biomechanics to determine force production, the way in which they were implemented means that impact and sustained force of the shoulders was quantified. It is, therefore, difficult to directly relate this data to spinal loading. The more recent live scrummaging study (Cazzola et al. 2015) used instrumented shoulder pads to measure impact and sustained force and therefore is again measuring force transmitted through the shoulders and not that borne by the spine.. Although this is extremely valuable information and has resulted in changes to scrum engagement laws, it does not provide data about spinal loading.

With regards to scrummaging kinematics, the more recent studies investigating rugby scrummaging used high speed video (HSV) to explore the kinematics of each player (Preatoni et al. 2013; Cazzola et al. 2015). This provided data relating to each individual player, however, since only three points were marked on the trunk, it only allowed for whole trunk orientation and engagement velocity to be defined as opposed to the motion of specific segments. Moreover, these points were defined on the playing shirts of the players which means additional potential for error on top of the assumption that skin motion artefact above bony landmarks is negligible. There have also been other studies investigating scrummaging kinematics although these only provided data in two dimensions (Rodano and Pedotti 1988; Sayers et al. 2009). Both these studies provided a lateral view during machine scrummaging and defined multiple body segments including the ankle, shank, thigh, trunk and arms. This provided data relating to whole body kinematics but only in two dimensions meaning one whole part of the kinematic 'story' was not provided.

Most recently individual scrummaging was investigated to determine specific characteristics of scrummaging of the different playing positions. No statistical differences ($p>0.05$) were observed between playing position for forward force production but front rows were at a higher vertical height than second rows at peak force production ($p=0.028$). Front row players also had lower centre of pressure variation than second or back rows ($p=0.044$) and required less movement to produce their maximum force ($p=0.020$) (Green et al. 2015). Although this research provides data relating to individual playing positions and their contribution to the scrum, it is difficult to determine whether these characteristics will apply to full, machine scrummaging and fully contested live scrummaging. Therefore, this data cannot be carried directly over into a game situation as it does not consider all the factors associated with scrummaging and considers a much more controlled situation.

Thus, although these previous studies have provided some valuable information relating to scrummaging biomechanics, there has not been specific focus on the spine which is where a large proportion of injuries are known to occur. Therefore, there is significant scope for research into scrummaging spinal biomechanics and this thesis aims to begin to quantify some of these parameters.

2.4.2. Forward Force Production

Previous research has concentrated on instrumenting scrum machines to measure the forward force production of rugby union scrums. More recently however, there has been success in quantifying forces in the live scrum.

A summary of the key findings of these studies looking at engagement and sustained scrummaging forces is given in Table 5. The magnitude of engagement force produced varies from 4.43kN (high school) (Milburn 1990) to 16.5kN (International) (Preatoni et al. 2013) and the magnitude of sustained force production varies from 3.37kN (high school) (Milburn 1990) to 8.3kN (International) (Preatoni et al. 2013). There may be a number of reasons for this large difference in impact and sustained force. Firstly, the disparity in playing level is evident. A higher playing level will mean greater skill and thus, better coordinated action from the members of the scrum therefore increasing forward force production. Furthermore, international packs will be much greater in size, mass and strength, all contributing to the increased force production. Differences in size and strength are particularly prominent in this era with a vast amount of time and money spent on professional players to increase these attributes. Players have progressively increased in size and strength since rugby union became a professional sport in 1995 (Olds 2001; Sedeaud et al. 2013; Fuller et al. 2013) with players at higher levels of the game being heavier and taller than those at lower levels (Fontana et al. 2015). When normalised for pack weight, however, only International and Elite level teams produced significantly more force ($p<0.05$) than other playing levels indicating that it is likely to be the greater skill and coordinated muscular action that affects forward force production (Preatoni et al. 2013). Owing to numerous law changes regarding the scrum over the years and this change in player physique, scrummaging techniques have evolved which could also contribute to greater forward force production.

Table 5 – Mean (standard deviation where given) of forward force produced by a scrum against a scrum machine across various playing standards

Author(s)	Method	Playing Level	Engagement Force/kN	Sustained Force/kN
Milburn (1990b)	Kistler model 9281B force platforms	University	6.54	4.61
		High School	4.43	3.37
Quarrie and Wilson (2000)	Strain gauge force transducers incorporated in shoulder pads of scrum machine frame	Semi-Professional	10.85 (1.17)	7.17 (1.18)
Du Toit et al (2005)	Shoulder pads fitted with pressure pads and VHF transmitters	School U19	7.53 (0.39)	6.15 (0.32)
Preatoni et al (2013)	Strain gauge force transducers on each of 4 pusher arms of scrum machine (vertical and lateral forces) and load cell (compression)	International	16.5 (1.4)	8.3 (1.0)
		Elite	16.5 (1.4)	8.0 (0.7)
		Community	12.0 (1.6)	5.8 (0.4)
		Academy	11.7 (2.0)	5.9 (0.8)
		Women	8.7 (0.1)	4.8 (0.5)
		School	9.1 (3.2)	4.9 (1.3)
Wu et al (2007)	Custom built individual scrummaging machine with uni-axial load cell	National	$1.00^{*\ddagger}$; $1.00^{\dagger\ddagger}$	Not Invesigated

* - Parallel foot position; † - Non-parallel foot position; ‡ - Values given are for individual players

Some research has looked at variables that cause a change in forward force production. Wu et al (2007) used a custom-built individual scrum machine instrumented with a uni-axial load cell and reported that forward force production was most effective when the front row players engaged with the scrum machine at 40% body height. Other factors showing some correlation with scrummaging force include body mass, somatotype, maximum anaerobic power, and isokinetic knee strength. Body mass (r=0.54; 95% CI=0.27-0.73) and maximum anaerobic power (r=0.51; 95% CI=0.24-0.71) were the two variables that correlated most with individual scrummaging force (Quarrie and Wilson 2000).

As has previously been mentioned, more recent research has been performed investigating different engagement procedures during live scrummaging. Table 6 shows the results from this study using three different engagement techniques (Cazzola et al. 2015). This data was obtained by estimating force using pressure sensors encased within the shoulder pads of the front row players of one forward pack. Each front row player of one team had a pair of pressure sensors used to collect pressure distribution data at 500Hz. The sensors were trimmed to fit into bespoke sleeves and then attached to the shoulders of each of these three players.

The three techniques used were ‘crouch-touch-pause-engage’ (CTPE), ‘crouch-touch-set’ (CTS), and ‘crouch-touch-set’ with a pre-binding (PB) of the props. From the data it can be seen that there was a large reduction on peak impact force from the CTPE to the PB sequences. The impact force on engagement is a phase that has been known to cause injury (Wetzler et al. 1998) and therefore, by reducing force it may reduce the severity of injury when incorrect engagement occurs. Furthermore, no significant differences were observed ($p>0.05$) for sustained force for any of the three sequences indicating that changing the sequence and reducing the impact force has no bearing on a team’s ability to generate sustained forward force.

Table 6 - Mean (standard deviation) of forward force during live scrummaging using different engagement sequences

	‘Crouch-touch-pause-engage’	‘Crouch-touch-set’	‘Crouch-touch-set’ with pre-bind
Peak impact force (kN)	9.8 (2.7)	8.8 (2.2)	6.3 (1.6)
Sustained shove force (kN)	4.2 (1.4)	3.8 (1.4)	3.8 (1.2)

2.4.3. Lateral and Vertical Force Components

Whilst primarily seeking to drive forward and thus displace the opposition from the ball, players still experience forces in both the lateral and vertical directions during scrum machine-based scenarios (Table 7). These forces presented are components of the force vector. For the data presented, negative vertical force corresponds to downward force and negative lateral force corresponds to lateral force to the left. This was obtained using the same method as the forward force data previously presented; that is, through the use of a force platform, strain gauges or load cells. All studies identified an element of lateral force, albeit without consensus describing the direction of force (Milburn 1990; Preatoni et al. 2013). This data is likely, however, to represent a fraction of that experienced during live scrums, as the opposing team tries to implement tactics to ensure they win ball possession, perhaps trying to manoeuvre the scrum in a certain direction. Such actions will create asymmetric spinal motion and load, hence, likely be a major contributing factor to cervical spine degeneration (Scher 1983).

Table 7 – Lateral and Vertical Force Production of Various Playing Levels against Scrum Machine. Note: Rightwards directed lateral force is positive and upward vertical force is positive. All values quoted are mean (standard deviation where given)

Author(s)	Method	Playing Level	Lateral Force/kN	Vertical Force/kN
Preatoni et al (2013)	Strain gauge force transducers on each of 4 pusher arms on scrum machine	International	0.6 (0.5)	1.1 (1.3)
		Elite	0.6 (0.4)	0.7 (0.9)
		Community	0.1 (0.6)	0.0 (0.9)
		Academy	0.1 (0.3)	0.1 (0.6)
		Women	-0.1 (0.3)	0.0 (0.5)
		School	0.1 (0.3)	0.1 (0.9)
Milburn (1990b)	Kistler model 9281B force platforms	University	-0.73	-0.16
		High School	-1.51	0.61

Vertical force also varies between different playing levels and across the two studies. The variation in force magnitude between studies is likely to be attributable to differences in ability and strength, though may also be a consequence of different scrum machines allowing different degrees of freedom. The difference in force is likely to be because teams engage in different ways. If a team engages with the machine driving upwards, a larger vertical force will be recorded rather than if they engage with the machine with the intention of driving solely horizontally. During live scrummaging, opposition players will attempt to counteract any upward force through the exertion of a greater downwards force to avoid inferior performance. A mismatch in force magnitude may result in either the scrum collapsing or players popping out.

2.4.4. Front Row Biomechanics and Force Distribution

During scrummaging, all forward force is transmitted through the front row. Previous research has identified an initial peak on engagement with a scrum-machine, before a lower force is sustained over an extended time period (Table 8). Furthermore, this research looked at how the force production of the whole scrum was distributed across the front row during machine scrummaging. This was measured by individual strain gauges on the individual pusher arms of the scrum machine. The fact that the hooker is consistently exposed both to the greatest impact and sustained force would indicate this is a likely cause of injury to both the shoulders and the spine. It is easier to infer shoulder injuries as forces to this area have been measured in previous scrummaging studies. Inferring cervical spine injuries, however, is more difficult. The reason the cervical spine is likely to be susceptible to injury is that the heads of the front row will be in contact with the chest of their opposition. Thus, if the force vector of the opposition has a downwards component, the players will need to counteract this to avoid the scrum collapsing. During this, the front rows will have to drive upwards, forcing their heads into the chests of the opposition. To stabilise the head, greater cervical spine muscle activity will be required to avoid unwanted displacement during the impact and sustained shove phases. This increased muscle activity may have injurious consequences in the long term such as premature degeneration of the cervical spine.

Table 8 – Force distribution across front row players during engagement and sustained scrummaging

Author(s)	**Playing Level**	**Engagement Force/N**			**Sustained Force/N**		
		Tight-head	*Hooker*	*Loose-head*	*Tight-head*	*Hooker*	*Loose-head*
Milburn (1990b)	University	2090	3380	1070	1390	2070	1150
	High School	810	2120	1500	630	1650	1090
Du Toit et al (2005)	School U19	2549	2866	2111	2203	2063	1929

There have been other investigations into the force distribution across the front row although this was a secondary aim of the respective studies.

When investigating second row hip and crotch binding (Milburn 1987), it was found that the loose-head prop bears almost 60% of the sustained force, but the tight-head prop bears the majority of the force during the engagement. This would suggest an asymmetry of scrummaging force distribution during the two distinct phases of the scrum. Contrary to this, however, in another study of machine scrummaging, Milburn (1990) found that the hooker carried almost half the load on engagement. In terms of lateral force, there was a consistent leftward direction of force application throughout the members of the front row and this type of force has been suggested to be a major contributing factor to premature spinal degeneration (Scher 1983). Furthermore, crotch binding produced greater downward vertical force although this was statistically insignificant ($p>0.05$).Greater downward force is likely to increase the likelihood of scrum collapse but since this finding was insignificant, this cannot be said with certainty.

Du Toit (1993) assessed the isokinetic neck strength in the sagittal and coronal planes of all players of the scrum, but the results concerning the front row are particularly relevant to this thesis. Front row players had the least neck strength during flexion (196N), then lateral bending (338N) and finally the most strength in extension (359N). Since front row players are in a position of neck flexion, during normal scrummaging and, anecdotally, scrum collapse (Scher 1982), they are likely to resist motion through cervical extension. Thus, the result showing their strength in cervical extension is particularly important. Additionally, it has been suggested, in a number of studies, that the impact and sustained forces in machine scrummaging regularly exceed the threshold for injury of the cervical spine (Milburn 1994). Although this is true, the force cannot directly be applied to the neck and must be borne by other anatomical structures, such as the shoulders, otherwise neck injury would be even more prevalent in front row players. According to Milburn (1994), peak impact during scrummaging for the front row is 12 times greater than the peak strength of the neck during flexion. For sustained scrummaging, force on the front rows was 9.9 times greater than peak cervical strength during flexion. If the necks of front row players were subjected to these loads directly, there would be regular catastrophic injuries to the spine, which is not the case.

During machine scrummaging, Milburn (1994) identified the force contributions of the various scrum units. These contributions were 37% from the front row, 42% from the second row, 9% from the flankers and 12% from the number eight. This indicates that not only is all the force transmitted through the front row, but they also generate a large amount of forward force. The second rows are the main source of forward force production and the back row players, the flankers and number eight, have only a limited amount of influence. Interestingly, the addition of more players did little to affect the vertical and lateral forces that were produced. Lateral and vertical force production has been suggested to be an indication of inefficient technique and this finding suggests that it is the front row themselves that produce the unwanted forces in the vertical and lateral directions. It is possible that this is a tactical manoeuvre, but it is unclear as to whether that was the case in this study. Furthermore, the engagement force produced by all 8 players simultaneously was under 6,000N, but the force produced when totalling each player's individual pushing force exceeded 17,000N. The force attenuation is most likely a result of the inefficient transfer of force through the players of the scrum. For example, the number 8's pushing force must be transmitted through the second and front rows before being recorded by the instrumented scrum machine. Much of this force attenuation will result from the trunk musculature, associated ligaments and the ability of the intervertebral discs to dissipate force effectively (Torg and Ramsey-Emrhein 1997).

2.4.5. Scrummaging Kinematics

Kinematic analysis during machine, and more recently, live scrummaging has previously been performed, but each study has had its strengths and weaknesses. Rodano and Pedotti (1988) was the first study to investigate scrummaging kinematics with markers placed on the axis of rotation of the shoulder, hip, knee, ankle, iliac crest and 5th metatarsal head. This provided data relating to body position, during the impact and sustained push in two dimensions in the sagittal plane. Although this provided some useful data, with regards to scrummaging performance and how different body positions may result in better force production, only a two dimensional analysis was performed, meaning one degree of rotational freedom was not recorded.

This problem of being restricted to two-dimensional analysis is also evident in a study performed by Sayers et al (2009), where a kinematic analysis of high-performance props during machine scrummaging was undertaken in 3-man, 5-man and 8-man scrums. During the 3-man scrum it was found that the magnitude of trunk angles were lower than for the other two conditions. Additionally, when props had the aid of the second and back row players, their rate of leg extension was reduced. It was determined that 3- and 5-man scrums do not replicate full 8-man scrums with regards to the kinematics of the props. It was suggested that it may even have a negative effect. The strength of this analysis was that multiple body segments were defined including head, trunk, upper and lower arms, thigh, shank and foot. This provided details of whole body segmental kinematics, but there were some limitations to this. The analysis performed was restricted to two dimensions. Furthermore, the trunk was defined as one segment, which does not account for the motion of different spinal segments. This is one of the largest problems of this study as it does not provide the full picture of spinal ROM, even as one single segment.

The most recent studies of scrummaging kinematics are that of Preatoni et al (2013, 2014) during machine scrummaging and Cazzola et al (2015) during live scrummaging. Both studies utilised similar protocols with regards to kinematic analysis. Two cameras were placed laterally on either side of the scrum, with two directly above the scrum on an overhead gantry. The lateral cameras allowed sagittal motion of the props to be viewed and the overhead cameras allowed transverse motion of the trunks of all 8 players to be viewed. The overhead cameras provided detail, relating to every member of the scrum, and this is valuable to determine trunk orientation of the different playing positions and, potentially, how different trunk orientations can affect force production. The problem is, however, that the trunk was defined as one segment and therefore, no segmental analysis could be performed on the spine. Furthermore, the analysis of each set of cameras is limited to one dimension. The lateral cameras measured sagittal plane motion of the props and the overhead cameras measured transverse plane motion of all players. Although having kinematic data for all members of the scrum is useful, it does not provide any detail relating to segmental motion of the trunk/spine, which may prove to be an important contribution to our understanding of potential scrummaging injury mechanisms.

In summary, previous studies have provided kinematic analysis in a variety of scenarios, which have contributed to our understanding of trunk kinematics in particular, as well as whole body kinematics. The problem is, however, that no studies have provided a three-dimensional kinematic analysis of any specific player, thus, leaving a gap in our knowledge of scrummaging biomechanics.

2.4.6. Electromyography during Scrummaging

Some studies have previously used EMG to determine muscle activity during rugby scrummaging. Otsuka et al. (2010) and Sharp et al. (2014) monitored trunk and lower extremity muscle activation, Piscione & Gamet (2006) investigated cervical extensor muscle activity during individual scrummaging against a scrum machine and Cazzola et al (2015) investigated spinal muscle activity in simulated scrummaging. This section also gives an overview of some injury-related literature of front row players and thus, the significance of the findings of these EMG studies.

Otsuka et al. (2010) recruited 6 healthy male volunteers and monitored the muscle activity of a number of trunk and lower extremity muscles during both the static and pushing phases of the scrum. Muscle activity was normalised to maximum voluntary contraction (MVC) and expressed as a percentage. It was found that the transversus abdominus and lumbar multifidus significantly ($p<0.05$) increased their activity from the static to the sustained phase with an increase ranging from 15-40%. No significant changes were found for any of the lower extremity muscles however. This large increase in muscle activity indicates the use of these muscles to maintain good posture in an unstable environment. The authors suggest that this unstable environment is likely a cause of lower back pain in many players.

Sharp et al. (2014) investigated EMG activity of the trunk musculature and that of the upper leg as well as quantifying horizontal force production. No significant difference in activation patterns for any of the muscles investigated were present and all muscles investigated showed a large amount of activity before the engagement phase. Post-engagement saw the muscle activity drop for all investigated muscles to a much lower and more consistent level. For example, the lumbar erector spinae muscles showed 78 ± 42.1%

MVC pre-engagement which then reduced to ~25% after first contact. MVC for this muscle group was obtained by the participants executing an upper torso lift. The erector spinae muscles were also found to be significantly more active ($p<0.01$) than either the hamstrings or the quadriceps during the pre-engagement phase and also significantly more active ($p<0.01$) than the hamstrings after first contact.

Piscione & Gamet (2006) investigated the level of muscle activity present in players of two different playing levels. They monitored the isometric contraction activity of the cervical erector spinae. The average activation, relative to MVC, of the cervical erector spinae was 62% for the U21 level players and 60% for the university level players. This muscle activity related to a mean cervical extension force of 430N (SD=94) and 295N (SD=39) respectively ($p<0.05$). Despite the difference in extensor force production of the two playing levels, muscle activity was insignificantly different ($p>0.05$). This would indicate that, regardless of playing level, a large amount of muscle activity is required to stabilise the cervical column during scrummaging. This large amount of activity warrants further investigation for both machine scrummaging of the full pack and also attempting to try and quantify cervical spine loading during live scrummaging.

The final and most recent study by Cazzola et al (2015) investigated spinal muscle activity during simulated machine and live scrummaging across two engagement sequences. These sequences were 'crouch-touch-set' (CTS) and 'crouch-bind-set' (CBS). During the sustained pushing phase, live scrummaging generated higher activation of the erector spinae musculature than machine scrummaging. Furthermore, the largest peak in muscle activity was observed just prior to engagement. The authors suggest this, observed for the new CBS engagement sequence, may prepare the cervical spine by stiffening the joints before impact and that machine scrummaging does not adequately replicate a live scenario. Thus, live scrummaging should be practised regularly to improve neuromuscular activation strategies to help with resisting external loads.

Increased paraspinal muscle activity can cause increased compressive loading of the spine which may contribute to spinal degeneration. This activity results in the loss of intervertebral disc height and resultant increased load bearing on the neural arch and uncovertebral joints (Skrzypiec et al. 2007). With time, this may lead to the development of osteophytes (Kumaresan et al. 2001). These findings are based on cadaveric modelling studies, however in vivo studies using MRI scans of rugby players have similar findings (Berge et al. 1999; Castinel et al. 2010). A relatively high incidence of disc narrowing of 35% (Castinel et al. 2010) and 71% (Berge et al. 1999) has been observed in rugby players, particularly the players of the front row. Furthermore, observation of osteophyte formation (83%), apophyseal joint degeneration (74%) and degeneration of endplates (77%) in front row players all suggest that these players are exposed to high compressive forces over a prolonged period of time. Hogan et al. (2010) similarly found apophyseal joint degeneration in front row rugby players, compared to controls with a significant difference at the C2/3, C4/5 and C6/7 levels. A study of two individual cases of cervical degeneration in front row players by O'Brien (1996) supports these findings. Anterior and posterior osteophyte formation was observed as well as disc narrowing at the C5/6 level and/or C4/5 or C6/7. Broughton (1993) also found the formation of osteophytes at the C5/6 and C6/7 level of a hooker. It is these types of chronic injuries that are found in the rugby union front row and are likely to be a result of cumulative compressive forces over time associated with repeated scrummaging and increased paraspinal muscle activity.

2.4.7. Summary of Scrummaging Biomechanics Literature

The scrum has been a research topic of interest for a number of decades and many studies have sought to quantify biomechanical parameters of machine scrummaging (Milburn 1987; Quarrie and Wilson 2000; Preatoni et al. 2013; Sharp et al. 2014) with particular emphasis on force production. More recently Cazzola et al (2015) investigated live scrummaging biomechanics which brought about changes to the rules that govern the scrum. World Rugby, formerly the IRB, changed the engagement sequence and also put into effect rules relating to how the ball is fed into the scrum. These new laws encourage the hooker to compete for the ball with a genuine 'hook' rather than previously where the scrum would drive over the ball in order to secure possession by 'winning the hit'. This would suggest that hooker spinal kinematics should, in theory, change as they are not acting as a player who contributes to forward drive, but instead must compete for possession of the ball. From this change to scrummaging technique, it is possible to hypothesis a potentially more injurious scenario for the hooker as they are forced to actively 'hook' and, thus, certain segments of the spine may be in more complex positions and more susceptible to injury. Therefore, the investigation of hooker spinal kinematics and cervical loading is an important addition to the previous scrum-related research that has been performed.

The review of the literature identifies the need to investigate hooker spinal biomechanics and this thesis aims to do so. The research gap is that no studies have attempted to quantify loads borne by the spine and this thesis aims to do so through the correlation of force and EMG and using EMG as a means of predicting the load borne by the cervical region. Another gap that was identified was the paucity of research into scrummaging kinematics, particularly that of the spine, which is an area that is so frequently known to be injured during scrummaging. Furthermore, owing to the law change, there may be an additional risk of injury when competing for the ball owing to the more extreme positions the spine may be subjected to. Thus, this thesis aims to address these gaps in the literature in relation to three different engagement techniques; machine scrummaging, live scrummaging of the old engagement sequence (CTPE), and live scrummaging of the new engagement sequence (CBS).

2.5. Playing Surfaces

This section provides an introduction to artificial turf as well some brief history and its development. It then goes onto look at various different aspects of both grass and artificial turf including injury epidemiology amongst various different sports, player perception of artificial turf and some mechanical properties of both surfaces. Various different sports are included in the epidemiology section and not solely focussing on rugby as very little research has been performed with just rugby in mind. As such, a number of other sports are reviewed including football (soccer) and American football.

2.5.1. Introduction to Artificial Turf

Traditionally, rugby has been played on natural turf (grass), but this is not necessarily just for the sake of tradition. In good conditions, grass turf provides an excellent surface for the game to be played on. Foot-to-surface interaction, traction and deformation, amongst other characteristics, are all of an adequate standard for even the highest standards of the game. Grass, however, has its disadvantages. One of the main disadvantages is that it requires a strict maintenance regime to keep it to the highest possible standards fit for International and national level teams (FIFA 2006). This is not only time consuming but also extremely costly.

Furthermore, the frequency at which pitches are used for both training and matches means that the surface will be subject to wear and tear from general use. With scrummaging in mind, an inadequate surface may destabilise the scrum which is associated with scrum collapse. Pitch maintenance is essential, but with such frequency of use, it is difficult to keep the surface at the highest standards all year around (FIFA 2006). Unless the pitch remains unused for a period of time, damage will continue to worsen.

Another factor that needs to be considered when using grass turf is the differing climatic conditions of numerous rugby playing nations. The vast array of different climates means that grass pitches will vary from being an ideal surface to one that is completely inadequate and, potentially, even dangerous to play on (IRB 2003). Synthetic turfs are much more resistant to adverse weather/climate conditions (FIFA 2006), thus, making them an interesting and realistic possibility for widespread use in rugby. These different climatic conditions will affect scrum stability as players require adequate purchase with the turf in order to scrummage effectively.

Over the years, a number of different turfs have been suggested for rugby, which have included surfaces such as clay, shale and artificial grass (IRB 2003), but none of these surfaces met the required standard for the sport. More recently, however, a new generation of artificial turf, known as 3G (3rd generation), has ‘caught the eye’ of the rugby world. The reason for this is that it is perceived that it will improve game quality by providing a consistent playing surface standard which will thus encourage high-quality and faster paced rugby. Some professional teams such as Cardiff Blues, London Saracens and Newcastle Falcons use 3G turf for match play and many local club sides do so for training provided they have access to such facilities.

Third generation (3G/synthetic) surfaces consist of a layer of synthetic grass with an in fill of rubber crumb. Underneath this is a shock pad layer, then a binding layer which adheres to the levelling layer. The inclusion of a shock pad layer is important as it reduces the maintenance required. The reason for this is that a good quality shock pad means that the pile of the synthetic grass can be shorter, meaning less rubber crumb is required. Beneath the levelling layer there is a sub-base, which is made from an unbound, graded, loose-laid aggregate. This surface gives an inert, stable, easy draining and frost-free surface. The last layer is the tarmac, or similar surface, to which the whole 3G surface is constructed upon. There are usually drainage pipes that run through the tarmac that collect water that runs through the surface. This type of synthetic surface has been designed to more accurately replicate the mechanical response of grass turf (IRB 2003), thereby eradicating extenuating ball bounce and high injury prevalence associated with earlier generations. Injury prevalence is perceived to be higher on synthetic turf than grass turf, however, Ekstrand et al (2006) reported an insignificant difference relating to ankle sprain incidence. Foot, ankle and knee injuries, however, all had a greater – although statistically insignificant – prevalence during synthetic turf gameplay (Fuller et al. 2010). Indeed, no author has yet been able to identify a direct cause-effect relationship relating injury incidence and play on synthetic turf (Ekstrand et al. 2006; Fuller, Dick, Corlette, et al. 2007; Steffen et al. 2007).

2.5.2. Shoe-Surface Interaction and Surface Traction

Previous research has focussed on the shoe-surface interaction and traction properties of various playing surfaces. This research may provide some insight into the mechanical properties of the surface and therefore have relevance to both performance and injury potential for athletes.

At the shoe-surface interface, traction is created, particularly when rapidly decelerating or changing direction. Too much traction at this interface may have the potential to increase the likelihood of a non-contact ACL injury of the knee and also twisting of the knee and ankle (Lambson et al. 1996). Dry and hot conditions on grass will mean harder ground and is suggested to increase the likelihood of ACL injury (Stiles et al. 2009) particularly if incorrect footwear is worn. In addition to traction, hardness is another property that has been suggested to be related to injury incidence. Orchard (2002) suggests that modifying the playing surface is the key to providing athletes with a universal method for reducing shoe-surface traction and therefore reducing the relative risk of injuries relating to shoe-surface locking such as ACL injuries and twisting of the ankle and knee.

When comparing traction of 3G and grass, Blackburn, Nicol, and Walker (2005) showed that the traction coefficient and peak torque produced on 3G pitches was lower than on grass. Peak vertical loads and loading profiles were very similar between the two surfaces. This is another positive argument for the use of 3G pitches as lower traction and peak torque are both risk factors for injury. Contrary to this, however, a more recent study has shown that the traction coefficients of 3G and grass were almost identical with subjects adjusting accordingly for the surface being played on. Despite the traction coefficient not differing, there were some differences in traction properties of 3G and grass although these were not statistically significant. Regardless of this, the authors suggest that 3G pitches should not be considered potentially hazardous with regard to excessive translational traction (McGhie and Ettema 2013). Although lower traction is a desirable property, this is only true to a certain extent. If the traction at the shoe-surface interface is too low, the potential for injury increases as with high traction. Therefore, an intermediate value of traction is the most desirable property at the shoe-surface interface (Torg et al. 1974).

As well as surface traction, the type of shoe used is an important consideration for the type of surface being played on. It has been shown that shoes intended for use on grass produced the highest peak torques on 3G (Livesay et al. 2006). It is therefore important for players to consider shoe choice, as the wrong shoe choice may cause in increased potential for injury. In spite of this evidence, generalising the use of a particular type of shoe for a given surface is difficult as loads experienced by each player depend on the playing position. In rugby, wingers will likely experience higher torques and traction as they rapidly move direction when running with the ball whereas forwards, particularly the front row will not. Therefore footwear with higher traction may be beneficial, particularly for scrummaging where good traction with the playing surface is essential although excessive surface traction will place excessive strain on ligaments and tendons. Another reason it is difficult to provide definitive recommendations on footwear is that each individual will have unique anatomical structures and biomechanical characteristics (Tillman et al. 2002) meaning a particular shoe may be suitable for one player, but not another.

Finally, a preliminary study was recently undertaken in order to determine player-surface interaction and surface performance characteristics during rugby kicking and simulated scrummaging with some preliminary results obtained. Large vertical forces were observed during simulated scrummaging which were much greater in magnitude than anterior-posterior and lateral forces. The main aim of the study, however, was to evaluate the methods used and therefore, no conclusive observations were made from the preliminary data (Ferrandino et al. 2015).

In conclusion, there are many considerations when choosing playing surface and footwear. A playing surface that produces an intermediate level of traction is desirable to reduce the risk of ankle and knee injuries. The choice of shoe is of equal importance as the wrong choice could cause increased injury risk. These are all important points that need be considered when playing sport on different surfaces and, arguably more than others, rugby is a sport that needs to pay close attention to this given the physical and aggressive nature of the game.

2.5.3. Injury Potential and Epidemiology on Different Surfaces

Changing the playing surface of various sports has been relatively common in recent years but there are a number of considerations that need to be accounted for; particularly how the playing surface might affect the potential for injury. In a study of artificial turfs it was found that movements that involve rapid deceleration, potentially coupled with an acute change in direction will cause large moments about the knee. These moments are likely to load the knee joints which could increase the potential for injury. When testing on 3G pitches however, there was a general trend that sagittal plane loading of the knee joint was reduced but this was not statistically significant ($p>0.05$). This is, however, a positive outcome for the use of 3G pitches (Blackburn et al. 2004). In addition to these findings, another study suggests that increased impacts, altered joint movement patterns and changes in the frictional coefficient of the playing surface are all contributing factors to the prevalence of overuse injuries. This was found to be true for artificial surfaces used in tennis where there was significantly more ground reaction force produced on an artificial surface ($p<0.05$) than the baseline test surface that was used (Stiles and Dixon 2006). In addition to the previously mentioned parameters that are likely to increase injury potential, harder surfaces cause increased eccentric muscle activity (Richie et al. 1993) and also differences in resistance to sliding have been suggested to facilitate an observed increase injury rate but no statistical evidence is evident. Thus, despite the numerous parameters that may increase the potential for injury, a direct cause-effect relationship has not been established between increased use of artificial turf and any particular type of injury.

There is some suggestion that grass turf yields a lower number of injuries when compared to artificial turf. These studies cite the deformation of grass turf to be a beneficial factor. Grass turf may have better deformation characteristics but it is much more difficult to get uniform mechanical properties across the whole playing surface whereas this is much easier to obtain on an artificial surface. Moreover, traction on a grass turf football (soccer) pitch is a function of soil type, soil density, soil moisture content and shoe-surface interaction amongst other variables (Stiles et al. 2009; Canaway 1975). All these characteristics of grass turf are extremely variable depending on climate thus making the case for using artificial turf stronger. Artificial turf has much less variability in its mechanical properties (FIFA 2006) regardless of climate and this consistency in properties may be beneficial for scrummaging as it provides a uniform surface.

In hotter climates, it is more than likely that grass pitches will be hard and dry, thus making the use of 3G pitches a desirable yet costly option. In a study of runners, it was found that the maximum reaction force was greater on a hard surface compared to grass but this was not significantly different ($p>0.05$). There may, however, be biomechanical implications of this data as this slight increase in reaction force may cause injuries in the long term; that is, fatigue/chronic injuries. Running on hard services may increase mechanical loads past the biological limits of the joints and tendons but, alternatively, running on soft surfaces may rapidly fatigue muscles and lead to injury (Tillman et al. 2002).

As well as climate conditions, financial considerations are also of importance particularly for less wealthy countries where sport is popular. The construction of artificial surfaces is extremely costly but maintenance is minimal whereas grass turf maintenance costs are much higher. If maintenance cannot be consistent, the mechanical properties of the pitch will change potentially increasing injury risk. When testing grass turfs of different age, it was found that newer grass results in quicker angular motion and lower axial torque with shorter duration. Furthermore, in a drop test, newer grass resulted in higher peak values in the horizontal and vertical directions which may have implications for injury since different turfs will offer different benefits (Kent et al. 2011). When higher torques are generated on a playing surface, there is an increased potential for injury, particularly to joints (Andréasson et al. 1986). One study suggests that the maximum torque produced is on grass which may contribute to injury (Nigg and Yeadon 1987). Therefore, countries who may not be able to afford the high maintenance costs will have grass turf pitches with very different properties and thus injury potential.

It is extremely important that when using a 3G surface it is properly maintained with an appropriate amount of rubber crumb covering the whole surface. When 3G is coupled with a shock absorbing pad or properly filled with rubber crumb, the attenuation of lower limb impact forces is much greater thus reducing the risk of overloading of the joints (McGhie and Ettema 2013). The maintenance of the surface, whether 3G or grass, is extremely important. It has been reported that wet conditions, on either surface, were associated with an increased injury risk. Moreover, the mechanical compliance of a surface with regards to impact is extremely important in many sports when considering injury potential (Milburn and Barry 1998).

Artificial surfaces may not only influence injury potential but also the performance of athletes. This performance aspect has both a positive and negative effect. On artificial surfaces, team sports in particular are faster and therefore much more entertaining for the audience but, with this increased speed of gameplay, there is a suggestion that there is an increased number of accidental injuries from collisions (Dixon et al. 1999). This is particularly relevant in rugby as collisions are so frequent and so an even greater number may cause an increased injury rate.

2.5.4. Summary of Playing Surfaces Literature

Rugby is more frequently being played on artificial turf and therefore, the literature surrounding playing surface research is becoming increasingly important. The shoe-surface interface and traction properties of the playing surface are important to consider with authors suggesting an intermediate level of traction as the most desirable and shoe choice is important. With regards to potential injury, it is difficult to conclude with certainty whether artificial surfaces result in more injuries. There is no statistically significant evidence to suggest that artificial surfaces cause more injuries but soft tissue and lower limb injuries appear to be more frequent on this surface. With specific regards to scrummaging, artificial turf may provide a more consistent playing surface than grass as it not so greatly affected by adverse weather conditions. This is likely to be beneficial to players of the scrum as the stability of the scrum will be affected by the condition of the playing surface.

2.6. Summary of Literature

First and foremost, this thesis aims to establish a method to measure biomechanical parameters of the spine during full machine scrummaging and fully contested live training scrums. Through a review of the literature, it was determined that inertial sensors are the best option to evaluate spinal kinematics owing to the fact that no line of sight is required amongst other advantages. Furthermore, if a relationship can be established between force and EMG for each individual, it may be possible to begin to understand loading of the cervical spine during scrummaging. It was determined that it is necessary to develop individualised relationships as well as determine the specific nature of the relationship as there are conflicting reports in the literature.

Injury epidemiology literature has reported that the scrum is a high risk event and results in a large proportion of rugby injuries. Furthermore, the scrum causes the second biggest percentage of spinal injuries after the tackle although this percentage depends on the study under consideration. Of these spinal injuries resulting from the scrum, the front row, particularly the hooker, are most susceptible to sustain an injury. Therefore, it was the hooker that was chosen to be investigated for this thesis.

Previous scrummaging research has predominantly concentrated on determining forward force production whilst also looking at lateral and vertical force production in a variety of scenarios ranging from types of binding to engagement technique. There has been some research into muscle activity during scrummaging but all these studies considered either individual or simulated scrummaging. Moreover, it was found that no in-depth data has been collected for a specific player and spinal loading is yet to be quantified during this event. Therefore, this thesis aims to address this gap in the literature by performing a comprehensive analysis of spinal biomechanics of the rugby union hooker during a number of different scenarios. One such scenario is the recent change in scrummaging laws and another is the effect of the recent shift in playing surface as this may have an effect on scrum stability.

3. Methods

This chapter provides the methods used for all three experimental chapters. The method used for the preliminary in-field experimentation and the subsequent chapters are very similar and therefore any changes to the protocol are summarised in the relevant sub-sections of this chapter. The initial aim of this thesis was to establish a method to measure biomechanical parameters of the spine during both machine and live scrummaging.

3.1. Inertial Sensor Validation

This sub-section presents the method use to validate the kinematic analysis technique chosen in the review of the literature.

3.1.1. Instrumentation

Six 3AMG sensors and a high-precision rotary table were used to measure orientation of each axis (roll, pitch, and heading). Roll is defined as rotation about the x-axis, pitch as rotation about the y-axis and heading as rotation about the z-axis. Figure 9 shows this definition

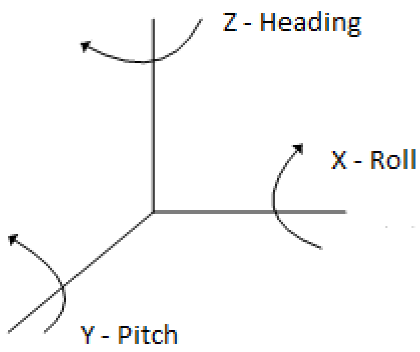

Figure 9 - Roll, pitch and heading definitions relative to their respective axes

The rotary table determined orientation through the use of digital encoders (ERN-420, Heidenhain, Sweden) each of which had 3600 lines per revolution with 4 steps per line giving steps of 1/40th of a degree. The lines were generated by etched marks which were optically scanned. Specifications of the digital encoders can be found in the manufacturer's guide (Heidenhain, 2013). The accuracy of the table was checked by rotating the table through 360° in each axis and checking the output of the digital encoders. The reason this method was chosen was that digital encoders are known to have a highly accurate measure of orientation to validate the chosen inertial sensor system against. Figure 10 shows the rotary table with digital encoders.

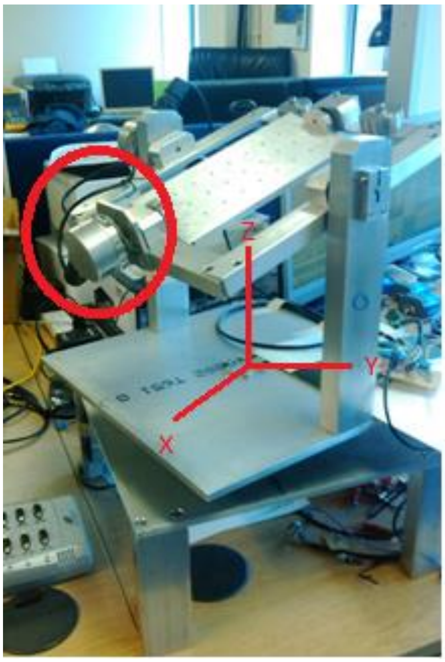

Figure 10 - Tri-axial rotary table used to validate inertial sensors with digital encoders used to measure orientation; red circle indicates location of digital encoder

3.1.2. Protocol

The six 3AMG sensors were attached to the top surface of the rotary table using double-sided adhesive tape. Double-sided adhesive tape is a widely used method of attachment in kinematic analysis along with Coban tape. In this scenario, double-sided adhesive tape provided a much more secure attachment than Coban tape. Each sensor was attached flush to a piece of metal and to each other so that all were in the same orientation. Figure 11 shows the setup of the 3AMG sensors during one of the trials.

Figure 11 - 3AMG set up on rotary table during validation trials. Trial shown is during pitch validation; red circle indicates location of 3AMG inertial sensors on rotary table

Having secured the sensors, two axes of the rotary table were manually secured at 0° orientation so no motion occurred in these axes during movement of the free axis. Once fixed, data collection for the rotary table and 3AMG sensors were started, and the table was rotated in the free axis from 0→+180°→0→-180°→0. This ROM was used as it covered the full range of the sensors' capability. This was done for three different speeds; slow (~10°/s), medium (~20°/s) and fast (~30°/s). A variety of speeds were chosen to validate the sensors as it was difficult to predict what angular velocities were to be expected during scrummaging. Although rotation speed was difficult to keep constant as the table had to be manually rotated, the use of a metronome helped keep this as constant as possible. Having completed these three trials for one axis, the table was fixed in two different axes and rotated through the same degree of motion for the free axis. This was repeated for the final free axis.

3.1.3. Data Processing

All data was exported to Matlab (Mathworks, 2012a) for processing. Both 3AMG and rotary table data were trimmed to start at the same point which was identified as where the first motion occurred; i.e. any deviation from 0°. If the data did not start from zero, it was magnitude-normalised to start at zero before being trimmed. There was very little noise in the data allowing for easy identification of the start point for each data set. The data was then time-normalised to 500 points so that both data sets had the same time base. The data was time normalised using a custom written Matlab code using linear interpolation which is a method of curve fitting using linear polynomials. This was because both pieces of equipment could not be synchronised and the sampling frequencies of the data sets was different. Data was presented in the form of a scatter plot with sample number along the x-axis and orientation (degrees) on the y-axis. As there was negligible wobble in the fixings (i.e. sensors securely fixed into place) the data was not filtered and left in its raw form.

3.1.4. Statistical Analysis

The absolute error was calculated for all data points to determine the mean absolute margin of error in orientation for each axis. An analysis of the data was performed (SPSS 18, SPSS Inc., Chicago, USA) to assess the concordance of the data between each sensor and the rotary table. This was performed using Lin's concordance correlation coefficient. This test was deemed to be more robust than correlation (e.g. Pearson's) as it provided a measure of how close the data sets were to a 1:1 relationship (i.e. deviation from 45° line).

3.2. Force-EMG Study Validation

This sub-section outlines the method developed for the analysis of force production of the cervical erector spinae musculature through a blinded study.

3.2.1. Instrumentation

A force platform (PS-2141, PASCO, California, USA) was secured in an inverted position, to measure force generation of the neck extensor muscles. Figure 12 shows a schematic of the force platform setup. On the left hand side, the diagram indicates the inverted position in which the platform was set up and the direction of force application from the participants. The right hand side schematic diagram shows a view of the underside of the force platform. The four corners are where the force transducers are located.

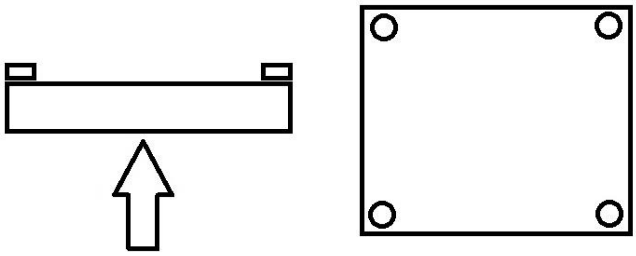

Figure 12 - Force platform setup. Left - force platform in an inverted position; arrow indicates direction of force application. Right - view of the underside of force platform; each circle represents a 'foot' of the force platform where force is measured.

EMG data was collected using a portable, wireless EMG system (PS850, Biometrics Ltd., UK). The system had a bandwidth of 20-450Hz and a common mode rejection ratio (CMRR) of 110dB at 60Hz. CMRR is the rejection of unwanted input signals common to both electrodes (inputs) relative to the desired difference signal. All raw signals collected by the system were pre-amplified with a gain of 1000, with data sampled at 1000Hz according to the Nyquist theorem to avoid aliasing. There is some debate about the sampling rate required for surface EMG, however, there is evidence to show that 1000Hz is sufficient and oversampling does not have a significant effect on the amplitude of processed signals (Ives and Wigglesworth 2003). SX230 bipolar EMG electrodes (Biometrics Ltd, UK) were used, with a fixed inter-electrode distance of 20mm. Electrodes were attached using double-sided adhesive tape (T350, Biometrics Ltd., UK) to unprepared skin as per the manufacturer's recommendation. A reference electrode was attached over the ulnar styloid using an elasticated wristband. Two electrodes were placed 10mm bilaterally at the C4/5 level, between the anterior margin of the trapezius and the midline of the muscle body (i.e. on the right and left cervical erector spinae (CES) in line with the muscle fibres) (Netto et al. 2007; Edmondston, Bjornsdottir, et al. 2011). The CES muscles are a prime contributor to neck extension although they are not the only muscle that aids neck extension (Netto et al. 2007). These particular muscles were chosen as they are a primary contributor to neck extension. The positions of the electrodes could be consistently identified through the palpation, initially, of the C7 spinous process. The C7 spinous process is particularly prominent on the neck but may be confused with the process of C6. To identify C7, both C6 and C7 were palpated and the participant was asked to extend their neck. The process of C6 glides away, while C7 remains prominent (Middleditch and Oliver 2005). The palpation of C7 provided a reliable reference point to then palpate upwards to the C4/5 level.

3.2.2. Experimental Procedure

Participants were instructed to perform a series of isometric neck muscle contractions by pushing upwards with the back of their head on the force platform, with gradually increasing contraction. Participants adopted a position of 90° hip flexion, placing their hands on the bench for support when performing contractions, but were instructed not to use their arms to aid force production. When performing the contractions, participants were asked to fix their eyes on a point marked on the floor to minimise the amount of sagittal plane cervical motion. Furthermore, participants were instructed to perform the contractions as if they were attempting to tilt their head to look upwards. This was done so that the muscles of interest were activated. Firstly, participants were allowed to warm up by performing a number of motions in each direction. They then practised contractions of gradually increasing force, characterised as minimum, 25% of maximum voluntary contraction (MVC), 50% MVC, 75% MVC and MVC (i.e. incremental contraction data). Exact representation of these contractions was not required, however, as it was a range of contractions from minimum to maximum which was of interest to create the correlation of force to muscle activity. Each contraction was held momentarily, before increasing to the next contraction, with the overall trial repeated three times. Participants were given two minutes rest in between each trial to allow for full physiological recovery (Burnett et al. 2007). This data served to explore the relationship between force and EMG, following which each participant performed self-randomised contractions (i.e. randomised contraction data), repeating each exertion three times. The researcher was blinded to the force output of the randomised trials. The reason for this was to minimise any possible bias and determine whether it is possible to predict muscular force production from EMG data alone having established an individualised relationship between force and EMG (i.e. researcher did not see actual force data until determined force values had been obtained). An individualised relationship is required as there are large variations in EMG amplitude between participants and therefore correlations cannot be transferred between participants (F Queisser et al. 1994).

3.2.3. Data Processing

All data was processed in Matlab (Mathworks, 2012a). The raw EMG data was demeaned, full wave rectified and low pass filtered using a zero-lag 4th order Butterworth filter with a cut-off frequency (f_c) of 5Hz to produce a linear envelope for each channel. The original, raw data can be seen in Figure 13.

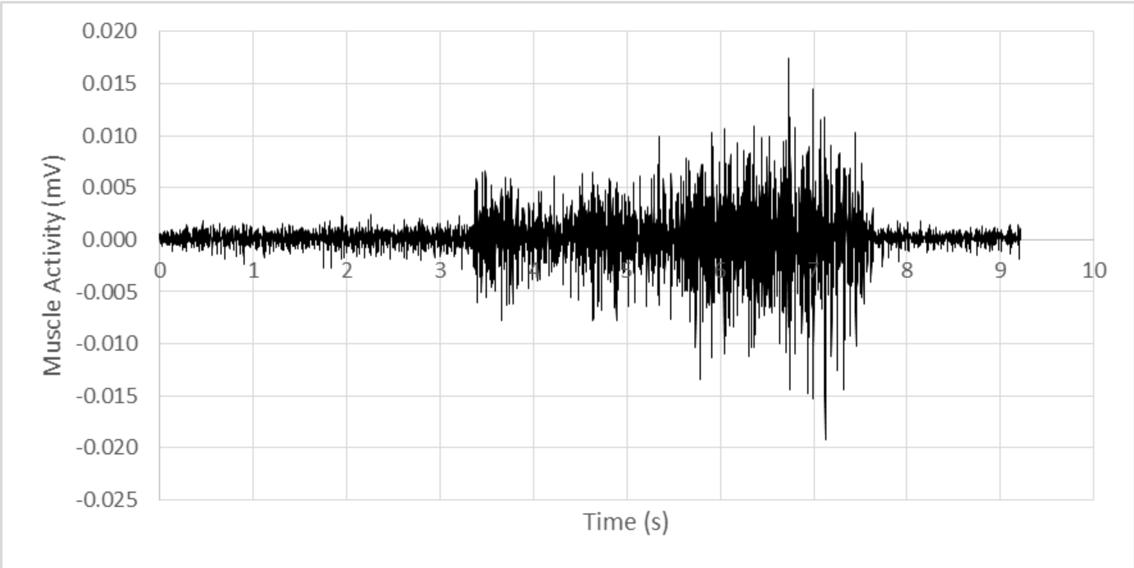

Figure 13 – Example of Raw, Unfiltered EMG Data

The data was demeaned in order to get rid of drift which is a standard processing step in EMG and the effect of demeaning the data can be seen in Figure 14. This process, however, did not yield any apparent changes to the data and therefore was not included in the subsequent analysis process.

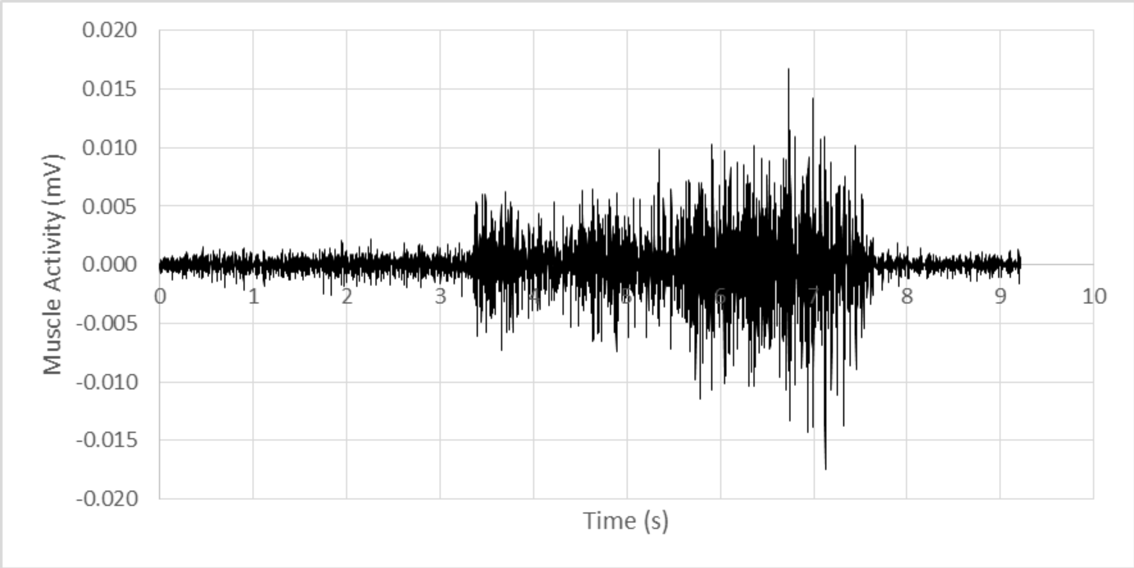

Figure 14 - Example of Demeaned EMG Data

The next processing step was to full wave rectify the data and the effect of this can be seen in Figure 15.

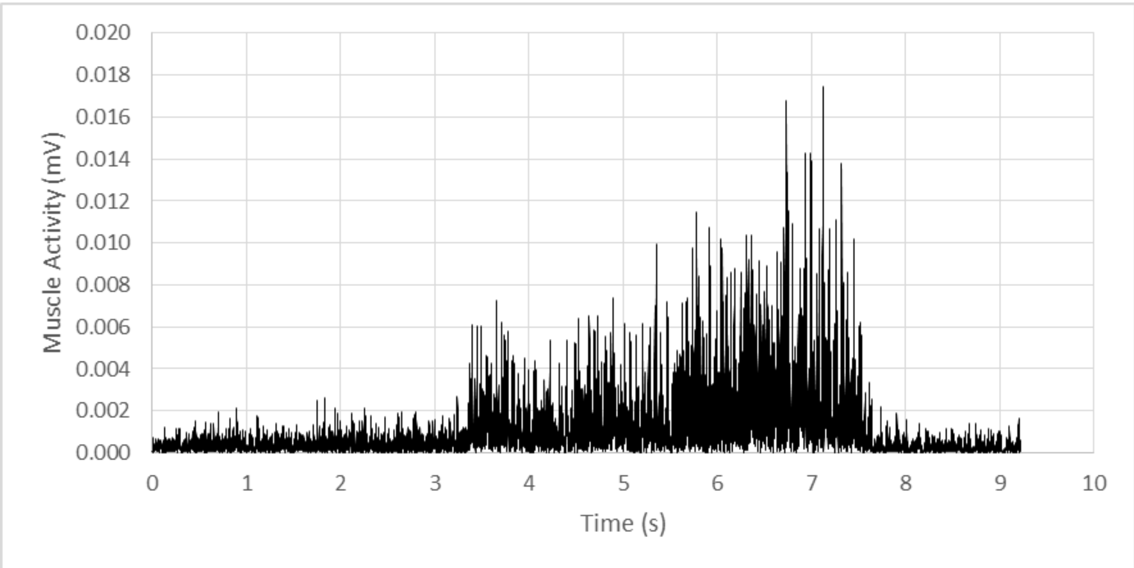

Figure 15 – Example of Rectified EMG Data

The final step was to low pass filter the data using a zero-lag 4th order Butterworth filter with a cut-off frequency of 5Hz (Netto and Burnett 2006; Burnett et al. 2008). The effect of low pass filtering the data can be seen in Figure 16.

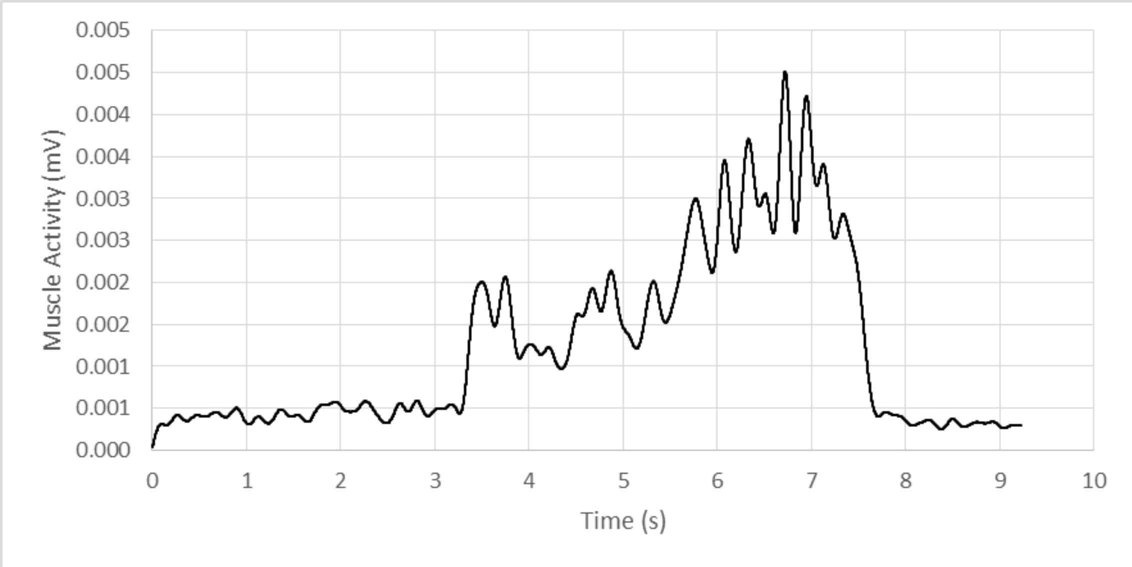

Figure 16 – Example of Final, Low-Pass Filtered EMG Data

The raw incremental force data was magnitude normalised in Matlab to start at zero. The five peaks in muscle activity and muscle force data for the incremental contraction data were identified and plotted. These peaks were identified manually. The largest peak in the force data was identified as well as the lesser peaks. The time intervals between these peaks were then calculated. The corresponding largest peak in EMG activity was identified and peaks at the same time intervals as the force data were identified. The largest peak in EMG activity was used as a reference. Given the linearity of the majority of such correlations (Nigg and Herzog 2007), a linear regression trend line was plotted for both EMG channels (left and right CES) versus muscular force using the method of least squares. Correlation coefficients (R^2) were calculated to determine the strength of the linearity of the muscle activity to force relationship.

3.2.4. Validation

The randomised EMG and force data were used to investigate the consistency and, thus, validity of the protocol. The randomised EMG data was processed as above with the addition of time normalising both EMG and force data. Muscular force was determined by "reading off" against the individualised correlation curves. The researcher remained blinded to the force data. Comparison between the determined and actual muscular forces was then performed, calculating the absolute and percentage errors for both left and right-sided EMG data. Left, right and an average of determined muscle force were then plotted against actual force for each participant.

3.2.5. Statistical Analysis

The Bland-Altman method was used to plot the difference between determined force and actual force for each participant, whilst the mean difference and upper and lower limits of agreement were also plotted. A two-tailed matched paired t-test was used to explore for significant differences between determined and actual force for all data points, and an analysis of the mean error was performed to determine the 95% confidence intervals (CI). A t-test was used as the aim of the study was to try and determine whether actual and determined force were statistically similar.

3.3. In-Field Testing Method

This section provides a detailed method that was used to evaluate spinal biomechanics of the rugby union hooker during machine and live scrummaging. It combines the two previous methodologies of kinematic and kinetic data collection in an in-situ scrummaging environment. It describes the instrumentation used, the data processing techniques and the trials in which the participants took part for all three experimental procedures. That is, the preliminary in-field trials, the playing surface study and the study investigating different engagement techniques.

3.3.1. Instrumentation

Inertial Sensors

A string of 6 inertial sensors (ThetaMetrix, Waterlooville, UK) were used to measure roll, pitch and heading data. The ability of the sensors allowed dynamic kinematic data to be collected for multiple spinal segments. Each inertial sensor had dimensions of L 42mm x W 30mm x H 12mm and a limit of ±2g. Accuracy of the sensors was ±5° in roll, ±4° in pitch and ±12° in heading during dynamic tests and sampled at 40Hz per sensor. The sensor string fed the information to the main processor unit (MPU) which in turn was connected to a laptop via USB. The software used was PearlSensors 3AMG (ThetaMetrix, Waterlooville, UK) and was custom built for the sensors. Data was saved as both .txt and .mat files to later be exported to Matlab (Mathworks, 2012a).

Force Platform

A uniaxial force platform (PS-2141, PASCO, California, USA) was attached in an inverted position to a scrummaging machine using a custom made rig (Palmer 2013) (Figure 17 & Figure 18). The red arrow indicates direction of force application in order to create individualised correlation curve. Blue arrows indicate direction of force as player engages with the machine and hits the pads with their shoulders. The thickness of the flat metal plate was 10mm and with a load of 150N applied upwards at each corner, the maximum deflection of the plate was 0.92mm (Palmer 2013). This force platform was selected as it provided the most cost effective solution for measuring vertical force production as required for this study. G-clamps were used to secure the force platform in an inverted position. The clamps were placed well away from the area of engagement so as not to interfere or injure the player. The red arrow indicates the direction of force application to activate the CES musculature in order to create the individualised correlation curves. This vertical force was measured by individual strain gauges at the corners of the force platform with the overall force applied being the sum of these four values. The blue arrows indicate the direction of force production during impact and sustained scrummaging with the shoulder pads of the scrum machine. The force platform was set up above the shoulder pads and therefore did not restrict movement of the participants. This is because participants' shoulders impact the pads with their head in between the pads. Therefore, the platform did not affect scrum position or restrict movement.

Figure 17 - Custom-made rig used to attach force platform to scrum machine from posterior view

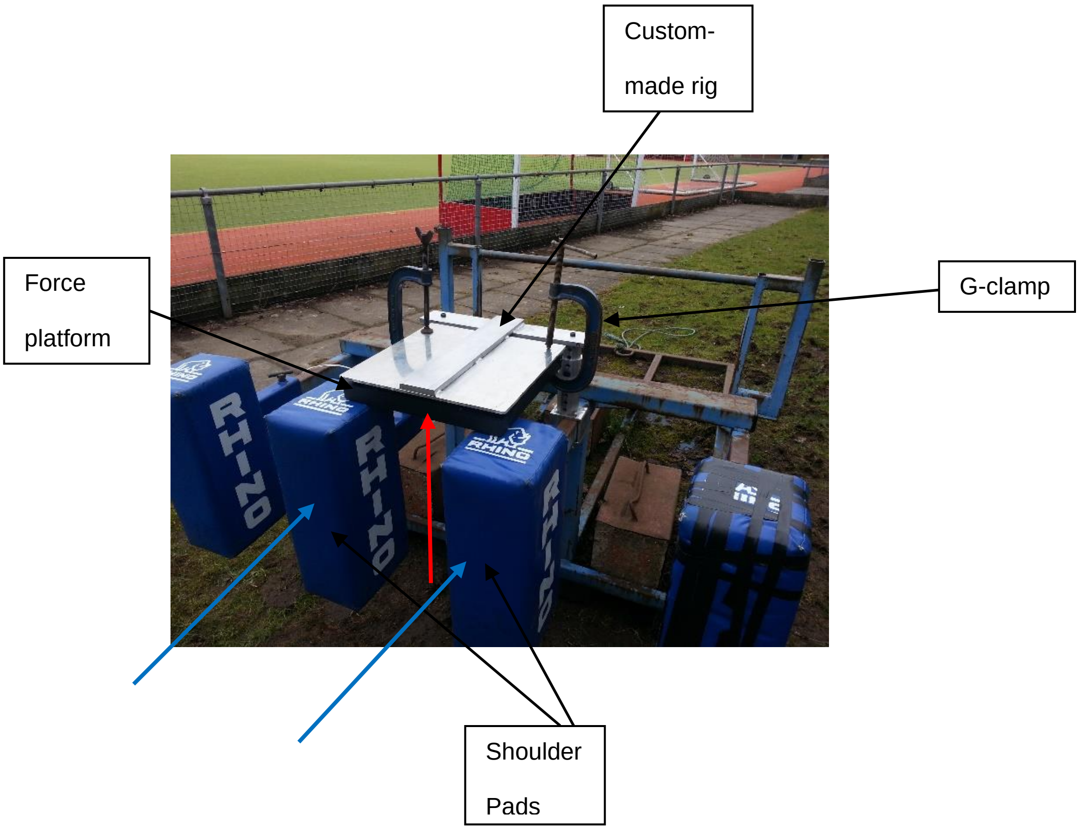

Figure 18 - Custom-made rig used to attach force platform to scrum machine in an inverted position with G-clamps.

The force platform was used to measure vertical force production during machine scrummaging in order to create an individualised correlation curve of force against muscle activity. The force platform was powered via USB and data was sampled at 1000Hz to match the sampling frequency of the EMG. When measuring force, a real-time plot of force against time was shown and the data was saved after each trial. Raw force data was saved in the standard SPARKvue experiment format (.spk file extension) and converted to a .txt file for post-processing in Matlab. The software used to collect force data was SPARKvue (PASCO, California, USA).

Electromyography Equipment – Preliminary Testing

A purpose-built EMG kit constructed at Cardiff University (Al Shaikh 2010) was used to measure muscle activity of the right and left cervical erector spinae (CES) muscles owing to the fact that the previously used kit was unavailable for use. The kit differed in that two individual electrodes had to be used to obtain the bipolar configuration rather than the electrodes being integrated into one unit. Furthermore, each muscle that was monitored required its own reference electrode rather than the single reference electrode used in the previous laboratory-based study. The wireless receiver unit was powered by the mains and two EMG channels were recorded; one for each muscle. The receiver unit was connected to the first two channels of a data translation box (Data Translation GmbH, Germany) which, in turn, was connected via USB to a laptop. The receiver unit wirelessly communicated with individual battery packs. The EMG electrodes (Neuroline 710, Ambu Ltd., Cambridge, UK) attached directly to the battery packs. Each pack had three inputs; one for the reference electrode and two to create the bipolar configuration required to monitor muscle activity. The reference electrodes were attached to the left and right sternoclavicular joints (Figure 19). According to EMG guidelines (SENIAM 2012) a reference electrode must be placed on a bony prominence as this is electrically inactive and therefore can be used as a 'reference' for the other electrodes. This reference location was chosen owing to the type of electrode used with this EMG equipment and was based on previous studies investigating similar musculature (Netto et al. 2007; K. Netto and Burnett 2006). Furthermore, the electrodes being used did not make it feasible to feed the electrodes down the leg and attach them to the proximal end of the tibia. The other four electrodes were attached 10mm bilaterally at the

C4/5 level between the anterior margin of the trapezius and the midline of the muscle body with an inter-electrode distance of 20mm on the left and right CES in line with the muscle fibres (Figure 19) (Netto et al. 2007; Edmondston, Bjornsdottir, et al. 2011).

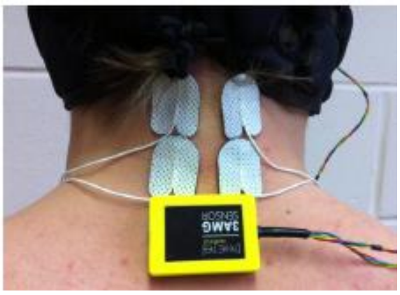

Figure 19 - Electrode positioning during preliminary experimentation. Four electrodes placed on left and right CES musculature (left) and reference electrodes placed on sternoclavicular joints (right)

All the electrodes were additionally secured with Hypafix tape (BSN Medical, Hamburg, Germany) to try and prevent sensors from falling off. All trailing wires were taped down so that they did not interfere when scrummaging. The battery packs were attached to the Velcro belt at the front of the participant so that they were out of the way of any binding. The software used was DT Chart Recorder (Data Translation GmbH, Germany) but this was later changed to QuickDAQ 2013 (Data Translation GmbH, Germany) owing to repeated malfunctions with the original software. All raw data was saved as .hpf files and later converted to a .txt file and imported into Matlab (Mathworks, 2012a) for data processing.

Electromyography Equipment – Different Engagement Techniques

Since some problems were encountered with the EMG equipment used in the preliminary in-field testing, a different set of equipment was used. The new equipment provided a better signal for a number of reasons: the inter-electrode distance was fixed avoid the problem of electrodes colliding; the elasticated band provided a secure fixation of the reference electrode and the data was saved on both an SD card and wirelessly transmitted to the computer to avoid any data loss.

Data was collected using a portable, wireless EMG system (PS850, Biometrics Ltd., UK). The system had a bandwidth of 20-450Hz and a common mode rejection ratio of 110dB at 60Hz. All raw signals collected by the system were pre-amplified with a gain of 1000, with data sampled at 1000Hz per channel according to the Nyquist theorem to avoid aliasing. SX230 bipolar EMG electrodes (Biometrics Ltd., UK) were used with a fixed inter-electrode distance of 20mm. Electrodes were attached using double-sided adhesive tape (T350, Biometrics Ltd., UK) to unprepared skin as per the manufacturer's recommendation. A reference electrode was attached over the proximal end of the tibia using an elasticated band. The position of the reference electrode was changed in this study as this EMG kit did not allow for easy adhesion to the sternoclavicular joint. Therefore, using an elasticated band around the proximal end of the tibia was more feasible and adhered to EMG guidelines (SENIAM 2012). The software used was DataLog which recorded in an .hpf file format which was later converted to a .txt format to import into Matlab (Mathworks, 2012a) for data analysis.

3.3.2. Experimental Protocol

Informed Consent

At the start of the training session, participants being investigated were asked to sign a consent form inquiring whether they fully understood the study procedure, have asked all questions they wish to ask, stating that they agree to participate in the study, and their data will be used solely for research purposes. All the data was fully anonymised to provide confidential data handling. It was made explicit to the participants that their involvement in the study was entirely voluntary and they were free to withdraw from the study at any time without giving reason. A copy of the consent form can be found in the appendices. Having agreed to take part in the study, the participants took part in their regular warm up as part of the team's training.

Participant Details Forms

Having agreed to participate in the study, all participants completed a details form to obtain background information. During preliminary in-field testing, details taken included age, height, weight, neck circumference, shoulder circumference, chest circumference, playing experience and dominant side.

During subsequent studies investigating playing surface and engagement techniques, this form was expanded to include, training sessions per week, scrummages per week, surface used for scrummaging, playing level, injury history as well as some other background information. A copy of both these forms can be found in the appendices.

Anthropometric Measurements

Having signed the consent form, anthropometric measurements were taken from the participants. Height, neck, chest and shoulder circumference were taken using a measuring tape and weighing scales were used to take the subject’s weight.

Inertial Sensor Placements

The string of 6 sensors were used to define five spinal segments. From a review of the literature the cervical spine was the segment that was of greatest interest as this is the area that has been known to suffer from many chronic and acute injuries during scrummaging. Owing to the size of the sensors, however, the cervical spine had to be defined as one segment rather than being split into multiple segments. During preliminary in-field testing, the landmarks were the forehead and the spinous processes of C7, T4, T8, T12, and S1 creating the cervical segment, upper, middle and lower thoracic segments, and a lumbar segment. This, however, was changed for subsequent studies because it was found in the preliminary study that there was minimal motion of the thoracic segments. Peak-to-peak ROM for the thoracic segments was as little as 30°. It has also been shown in the literature that injuries to the thoracic spine are rare in rugby (Targett 1998) although other injuries to the thorax are more common (Hayashi et al. 2014). Furthermore, there was more motion observed in the lumbar spine (up to 40°) and therefore, segmenting this region was of more interest. In addition to this, the lumbar spine is susceptible to injury from scrummaging (C W Fuller, Brooks and Kemp 2007). The landmarks used in the other studies were the forehead and the spinous processes of C7, T7, T12, L3 and S1 creating the cervical segment, the upper and lower thoracic segments and the upper and lower lumbar segments.

The sensors were attached directly to the skin using double-sided hypoallergenic tape. In addition to this, Hypafix tape (BSN Medical, Hamburg, Germany), was used to secure the inertial sensors in place. This tape was also used to tape down any loose wires. Any excess wire was taped down to the participant’s skin to minimise movement of the sensors and avoid any movement during binding of the props. A Velcro belt was wrapped around the participant's waist to attach the MPU to. The MPU was placed on the front of the participant so that it did not interfere with any binding from other participants. Figure 20 shows the positions of the inertial sensors over the spinous processes of one of the participants during preliminary in-field testing.

Figure 20 - Inertial Sensor Placements over the spinous processes of C7, T4, T8, T12 and S1

The inertial sensors were placed so that the rotations of roll, pitch and heading represented flexion-extension, axial rotation and lateral bending respectively when in a standing position.

Figure 21 shows the positions of the inertial sensors over the spinous processes of one of the participants during the playing surface and engagement technique studies. The figure also shows the orientation of the sensors having been changed. This was to allow for better adhesion to the skin of the players as a greater amount of the sensors' footprint was in contact with the skin. Therefore, the inertial sensors were placed so that the rotations of pitch, roll, and heading represented flexion-extension, axial rotation and lateral bending respectively when in a standing position.

Figure 21 - Inertial sensor placements over the spinous processes of C7, T7, T12, L3 and S1

Palpation of Vertebrae and Muscles

All spinal landmarks were found through palpation. Once a particular landmark was identified, a small mark was made on the skin with a pen so that the sensor could be attached to the skin above the landmark, once all landmarks had been palpated. The C7 spinous process is particularly prominent on the neck, but may be confused with the process of C6. To identify C7, both C6 and C7 were palpated and the participant was asked to extend their neck. The process of C6 glides away, while C7 remains prominent (Middleditch and Oliver 2005). T4 was found by counting down the spinous processes from C7. T8 was identified from the identification of T7. T7 was found by finding the inferior borders of the scapulae and finding the midpoint of a line drawn in the transverse plane connecting these points (Willems et al. 1996). T12 was found by counting up from the palpation of L4. L4 was identified by a line bisecting in the transverse plane at the most superior point of the iliac crests (Burton 1986). Similarly, for the latter two studies, L3 was found by counting up from L4. The spinous process of S1 was found by finding the midpoint of a line in the transverse plane created by the posterior superior iliac spines (Chakraverty et al. 2007).

In order to aid finding these landmarks, the researcher was provided some informal training from a registered physiotherapist with expertise in spinal biomechanics. An initial training session was organised in order to demonstrate the palpation techniques and then the researcher used a variety of volunteers to practise these techniques on at a later date. A second session was organised with the physiotherapist in order to assess the repeatability and accuracy of finding the specific spinal landmarks. Furthermore, recommended texts were provided from the physiotherapist in order to help with this particular aspect of the project.

Range of Motion

Before attaching all the EMG equipment, if required for the study, the participant performed a series of trials to characterise their normal spinal range of motion (ROM). The movements were firstly explained to the participants, before allowing the participants to familiarise themselves with them. The ROM trials were all recorded in one trial, but were subdivided into three main sections; standing spinal ROM, standing cervical ROM and scrummaging cervical ROM. Between each section, the data was ‘time marked’ to make identification of each of these points easier when data processing. Each motion in a particular plane was performed three times with a short pause in between each before moving onto the next motion. The participant always resumed a neutral position between movements. The trial was performed in a pre-defined order. Firstly, from a neutral standing position, the participant performed full spinal flexion and then extension three times in total. Then they moved onto right and left lateral bending and finally right and left rotation. The data was then time marked and the participant performed the same series of movements in the same order, but just for the cervical spine. Having completed this part of the trial, the participant performed the same cervical ROM trials adopting 90° hip flexion, similar to a scrummaging position.

Machine Scrummaging

Having completed the ROM trials, the participant was asked to scrummage against an instrumented scrum machine as part of a full 8-man scrum if the scrum machine was available and the rig could be attached to the machine. If the rig could not be attached then just kinematic data was recorded. The machine scrummaging trials were performed for the preliminary testing and the study investigating engagement techniques. A custom made rig (Figure 18) was attached to the machine and G-clamps were used to attach a force platform in an inverted position (PS-2141, PASCO, California, USA). The participant was then asked to scrummage against the machine as normal with the back of their head pushing up against the inverted force platform. The participant was instructed to push with increasing exertion characterised as minimum, 25% maximum voluntary contraction (MVC), 50% MVC, 75% MVC and MVC as presented in the force-EMG study method (Section 3.1.2.2.). This process was repeated two more times. The kinematic data was sampled at 40Hz/sensor as this was the maximum the USB port could support and the force platform and EMG data was sampled at 1000Hz to allow for simpler synchronisation of these 2 data sets. The EMG and force data from these trials were used to create an individualised correlation curve of force and EMG, which was then used as a means of predicting force during live scrummaging based on the previously presented method (Section 3.1.2.). To keep the scenario as consistent as possible, a number of variables were controlled to the best of the researcher's ability. Firstly, the same 8 players were used to engage with the scrum machine for all five machine scrummaging trials. The participants also always scrummaged as the attacking team. This meant that the ball was always fed in on the left hand side of the scrum. Furthermore, the same scrum half was used and also the same coach was used to call the engagement sequence for each training session. During preliminary testing the engagement call used was 'crouch-touch-pause-engage' as these were the laws being used at the time of the investigation. During the study of different engagement techniques, the call used was 'crouch-bind-set', Keeping all these variables constant as best as possible, meant that the scenario in which the hooker scrummaged was kept as constant as possible.

Live Scrummaging

Having completed the machine scrummaging trials, the participants were asked to take part in a series of live scrummaging trials in a competitive, training environment with the inertial sensors and EMG electrodes attached. Each participant took part in five live scrums using the 'crouch-touch-pause-engage' sequence. This was during preliminary testing and therefore the study was conducted during the 2012-13 playing season. Thus, the 2014 scrum engagements laws had not been brought into play at the time of the investigation. During the study of playing surfaces and different engagement techniques, two different scrummaging sequences were used. Three scrummages were using the old sequence of 'crouch-touch-pause-engage' and three were using the new sequence of 'crouch-bind-set'. The order in which the engagement techniques were performed was randomised. Prior to testing, participants were assigned a value, via a table of randomised numbers, as to the order they would perform the two engagement techniques. This was done to reduce the possibility of any confounding factors. For the participants that took part in live scrummaging, they performed these trials on the surface available to their team. This was either grass or synthetic (3G) turf. All of the participants that took part in the machine scrummaging trials performed live scrummaging of both engagement sequences during live scrummaging on grass. The remaining 15 participants took part in the live scrummaging trials of the surface available to their team (grass or 3G).

The engagement sequence was called by the coach in charge of the training session. As many external variables were controlled, to try and keep the testing conditions as similar as possible. Variables that were kept constant were that the hooker was always on the attacking side which mean that ball was always fed into the left-hand side of the scrum. Furthermore, the scrum half was kept the same during each trial so that the way in which they feed the ball into the scrum was kept as constant as possible. The same coach was used to call the engagement sequence so that the intervals between each section of the engagement call were kept as constant as possible. Finally, the members of each scrum were kept the same during each testing session.

3.4. Data Processing

3.4.1. Electromyography

This processing was only carried out for the preliminary study and study of different engagement techniques as it was during these studies that data was collected. All raw data was converted to .txt files and imported into Matlab (Mathworks, 2012a) for data processing. The raw data was full wave rectified and low pass filtered using a zero-lag 4th order Butterworth filter, with a cut-off frequency (f_c) of 5Hz to produce a linear envelope for each EMG channel (Netto et al. 2007; K. Netto and Burnett 2006; Burnett et al. 2007). This previous research investigated similar musculature and thus, processing techniques were adopted from this research. These steps are standard processing methods for EMG data and are well established (SENIAM 2012). Furthermore, the effect of each processing step can be found in Section 3.1.2.3.

The average static muscle activity was calculated for both left and right sides for the first 600 samples (0.6s). The reason for this was so that the onset of muscle activity could be determined. The onset of muscle activity was defined as the muscle showing greater than three standard deviations either side of the average static muscle activity for a period of 30 samples (0.03s) (Sasaki et al. 2014). The EMG data was normalised to 10,000 points to match the number of samples in the force data.

3.4.2. Force Platform

The raw force data was converted to .txt files for each participant from their raw format of .spk files. Text files were imported into Matlab and processed using a custom-written Matlab script. In order to ensure a secure attachment of the force platform to the rig, the use of the clamps created some force on the platform despite there being no force applied by the participant. When processing, this value was subtracted from the trace so that the data started at zero. The force data was 'time normalised' to 10,000 points through the process of linear interpolation, similar to the EMG data. This allowed for easier identification of the peaks and other peaks in the data at the same time points.

3.4.3. Force-EMG Correlation Curve

To create the force-EMG correlation curve, processed force and EMG data from the machine scrummaging trials were used. The five peaks in muscle activity and muscle force data for the incremental contraction data were identified and plotted. These peaks were identified manually. The largest peak in the force data was identified as well as the lesser peaks. The time intervals between these peaks were then calculated. The corresponding largest peak in EMG activity was identified and peaks at the same time intervals as the force data were identified. The largest peak in EMG activity was used as a reference. This was done for each of the three gradually increasing exertion trials and the data was then collated to create a curve of 15 points in total with force on the y-axis and muscle activity on the x-axis. For the muscles being investigated, there has been some disagreement as to whether the force-EMG relationship is linear (Keshner et al. 1989; Queisser et al. 1994) or non-linear (Schüldt and Harms-Ringdahl 1988) so a number of different curves were fitted to the data. It was found that a linear fit resulted in the best correlation (R^2) values and so this relationship was used. This individualised correlation curve was used later on to determine neck muscle vertical force production during live scrummaging.

3.4.4. Inertial Sensors

The inertial sensors' software created .mat files and so this data was imported directly into Matlab (Mathworks, 2012a). Resultant angles between adjacent sensors were calculated to determine three dimensional motion for each segment through the use of a rotation matrix (Lee et al. 2003; Williams et al. 2013a). A rotation matrix is a 3x3 matrix where each 3x1 column is a unit vector representing the orientation of the sensor relative to the sensor's global coordinate system. It is a method of defining an angle unambiguously.

This analysis performed through a custom written Matlab code (Mathworks, 2012a) in the order of pitch, heading and roll for the preliminary study and pitch, roll and heading for the subsequent studies. This was changed owing to the change in sensor orientation. Motion data was filtered using a bi-directional zero-lag 4^{th} order Butterworth filter with a cut-off frequency of 6Hz (Fioretti 1996). Higher cut-off frequencies were used to filter the data and did not have an effect on the data. Therefore, the standard of 6Hz during motion analysis was used. Having removed the high frequency noise, the 5-point differentiation method was used to calculate angular velocity. This method was chosen over the 3-point method to yield greater accuracy when calculating angular velocity. Peak and mean range of motion and angular velocity were calculated. Positive angular velocity relates to motion in the direction of flexion, right lateral bending and right rotation. Negative angular velocity relates to motion in the direction of extension, left lateral bending and left rotation. Using rotation as an example, positive angular velocity meant movement in the direction of right rotation of a spinal segment and negative values meant movement in the direction away from right rotation (i.e. left rotation) of a spinal segment. The neutral position was defined as the standing position adopted at the start of the ROM trials, serving as the reference plane for all subsequent data; hence, a position of 30º upper lumbar flexion is 30º relative to the standing position.

3.4.5. Statistical Analysis

For each of the studies, the correlation coefficient was calculated to determine whether dominant side had any effect on peak normal active ROM. Furthermore, Pearson's correlation coefficient was calculated to determine whether, for example, number of training sessions per week affected peak active ROM.

To explore any significant differences between machine and live scrummaging peak kinematic data in the preliminary study, a matched pairs t-test was applied with significance set at $p<0.05$. This test was chosen having tested for data normality. Furthermore, a two-tailed matched pairs t-test was used to explore significant differences in peak EMG and peak force data. The null hypothesis is that no significant differences will be observed between machine and live scrummaging for any of the variables measured.

For the study of playing surfaces, a two-tailed independent t-test (with Bonferroni correction) was used was used to determine which motions or velocities demonstrated a significant difference ($p<0.05$). An independent t-test was chosen as the cohort was split into two independent groups depending on pitch availability. The null hypothesis is that no significant differences will be observed for spinal kinematics between playing surfaces. Furthermore, this was done for both the engagement sequences with the variable condition being the playing surface with all other variables being kept consistent. Furthermore, Cohen's effect size (d) was calculated to determine the magnitude of differences between conditions and $d>0.8$ was considered to be a 'large' effect. The analysis for the CTPE engagement sequence is given in the appendix. The CBS analysis is presented in Chapter 5.

In the final study, for the participants who took part in machine (CBS), live (CBS) and live (CTPE) scrummaging trials, a one-way repeated measures analysis of variance (with scrummaging condition as the within-group factor) was applied. This was performed to test for possible changes in kinematic, peak muscle activity and peak (determined) force across the three conditions, which was followed by Bonferroni post-hoc comparisons

($p<0.05$). The sphericity of the data was checked by applying Mauchly's test. All statistical tests were performed in SPSS (SPSS 22, SPSS Inc., Chicago, USA.).

Furthermore, for all three experimental chapters, the coefficient of variation (CoV) was determined for peak-to-peak ROM for each spinal segment in each anatomical plane of motion. The coefficient of variation is defined as the standard deviation of the data divided by the mean. In this thesis, for example, the peak-to-peak cervical sagittal ROM was determined for a particular participant for all trials for a participant condition (e.g. machine scrummaging). The mean peak-to-peak and standard deviation was then calculated and the CoV calculated. Finally, the CoV values were collated for each different condition (e.g. machine vs live scrummaging) and an ANOVA was used to determine whether the variation between conditions was significant ($p<0.05$). The null hypothesis is that no significant differences will be observed for peak-to-peak spinal ROM between any scrummaging conditions.

4. Preliminary Results

4.1. Introduction

Biomechanics consists of both kinematic and kinetic parameters and thus, a method that can measure both concurrently, will provide valuable data relating to spinal biomechanics during scrummaging. Owing to the paucity of research regarding spinal biomechanics during machine and live scrummaging, a method needed to be developed in order to meet this aim.

Spinal kinematic assessment is essential in both clinical environments and biomechanical research, in areas such as rehabilitation, injury biomechanics and sports biomechanics. A knowledge of spinal kinematics allows researchers to obtain vital information on injury mechanisms. Inertial sensors have many uses in kinematic studies and their use varies from clinical settings to biomechanics research including the assessment of human movement (Boonstra et al. 2006; Saber-Sheikh et al. 2010) and spinal biomechanics (Wong and Wong 2008). As outlined in the review of the literature, inertial sensors were chosen as they were deemed to be the most suitable method for kinematic data collection given the criteria for this particular application.

The other strand of biomechanical analysis is kinetics. This is most frequently conducted in the laboratory using force plates to measure ground reaction force (De Wit et al. 2000); however their use is rare in clinical and sporting environments. Another possible method of indirectly measuring force is through the use of electromyography (EMG). By assessing muscle force production through monitoring the electrical activity of specific muscles, information can be provided which helps to determine both muscle health and the loading tolerance of joints and soft tissues. One viable option for quantifying muscular force may be to use electromyography (EMG) which measures the electrical activity of muscle. EMG has various applications in the clinical setting (Williams et al. 2013a; Dolan et al. 2001), the research environment (Netto et al. 2007), ergonomics research (Edmondston, Sharp, et al. 2011) and even the sporting environment (Piscione and Gamet 2006; Sharp et al. 2014; Cazzola et al. 2015). If a relationship can be established between force and EMG, then it may prove possible to use EMG alone to predict muscle force, an approach which has been suggested for the lumbar spine (Dolan and Adams 1993; Dolan et al. 1999). Such a method could then offer a solution to the measurement of force when traditional load cell or force plate methods are inappropriate thus, removing environmental and technical constraints.

With a combination of both kinematic and kinetic data collection methods, it may be possible to investigate spinal biomechanics during rugby scrummaging in both machine and live environments.

4.2. Aims and Objectives

The overall aim of this chapter was to develop a method to measure kinematics and kinetics of the hooker's spine simultaneously during machine and live scrummaging.

4.2.1. Inertial Sensors Validation Aims and Objectives

The aim of this study was to validate a string of six inertial sensors (3AMG, ThetaMetrix, Waterlooville, UK) against a high precision rotary table to determine their accuracy for orientation data collection. The method for this study can be found in the previous chapter (Section 3.1.1).

4.2.2. EMG-Force Study Aims and Objectives

This study aims to validate a method to determine whether it is possible to predict the muscle force production of the cervical erector spinae during randomised contractions using individualised correlation curves of force and EMG. The method for this study can be found in the previous chapter (Section 3.1.2).

4.2.3. Pilot In-Field Study Aims and Objectives

The aim of this study was to merge the kinematic and kinetic methodologies to quantify biomechanical parameters of the spine of the rugby union hooker. This included kinematic data for five spinal segments as well as muscle activity and force during machine scrummaging and muscle activity and determined force during live scrummaging. The method for this study can be found in the previous chapter (Section 3.1.3).

4.3. Results

4.3.1. Inertial Sensor Validation Results

This section presents results of a specific sensor relative to the rotary table for all three axes during the fast rotation trials. The figures for all the other sensors can be found in the appendices (Figures

A.1-A.15). Figure 22 shows motion in roll of the first inertial sensor against the corresponding motion of the rotary table. Through 0 to +180° there is very little deviation from the 45° line of best fit but there is some deviation of up to 10° for the lower range of 0 to -180°. It should be noted, that the sensors were utilised from -60° to +60° for in-field testing but were validated for their full range during laboratory experimentation.

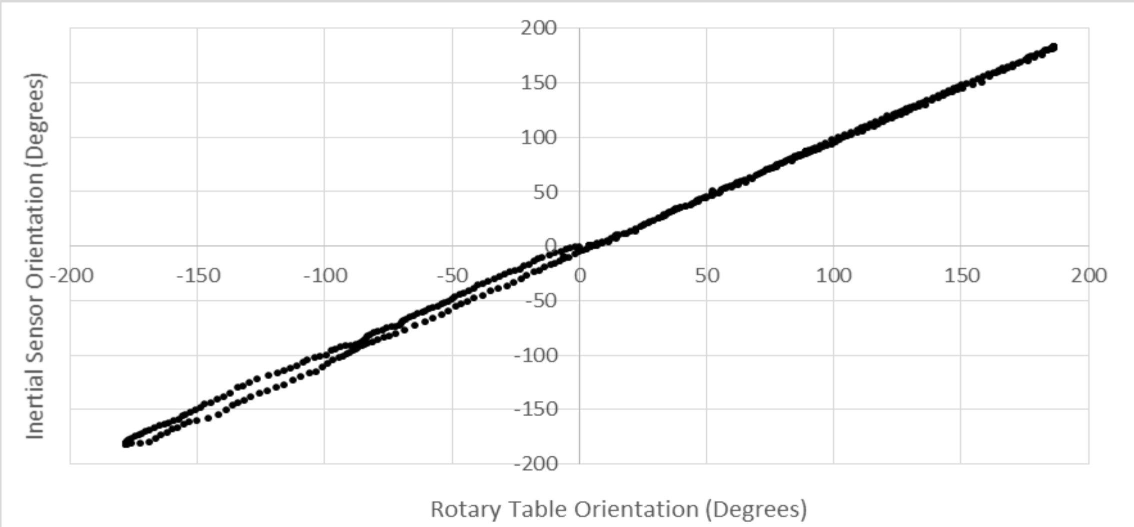

Figure 22 - Rotary table orientation (x-axis) against inertial sensor 1 orientation (y-axis) for roll

Figure 23 shows motion in pitch of the first inertial sensor against the corresponding motion of the rotary table. Through the full range of the sensor's capability (-90° to +90°), there is some difference between the two measurement techniques. The data points lie with little deviation from the ideal line of best fit at 45° to the horizontal. The maximum difference between measurement techniques is ~4°.

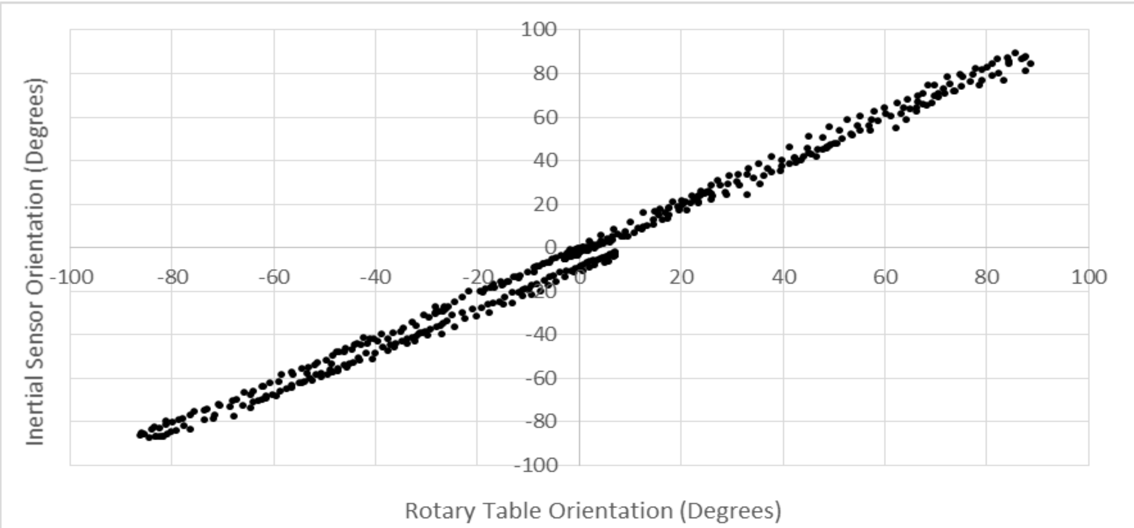

Figure 23 - Rotary table orientation (x-axis) against inertial sensor 1 orientation (y-axis) for pitch

Figure 24 shows motion in heading of the first inertial sensor against the corresponding motion of the rotary table. Through the full range of the sensor's capability (-180° to 180°), there are some differences between the measurement techniques. These are particularly evident when moving from +50° to the of its range. This particular pattern shown in this plot is also evident for all the sensors' heading axis and can be seen in the appendix (Figures A.1-A.15). This may be owing to the fact that this data is returned by a fusion from all the sensing elements unlike these previous two axes which are returned by a combination of the accelerometers and gyroscopes. The inclusion of the magnetometer is likely to increase the error as this particular sensing element is affected by the presence of ferrous metals. Furthermore, magnetic dip is likely to have affected these results (de Vries et al. 2009).

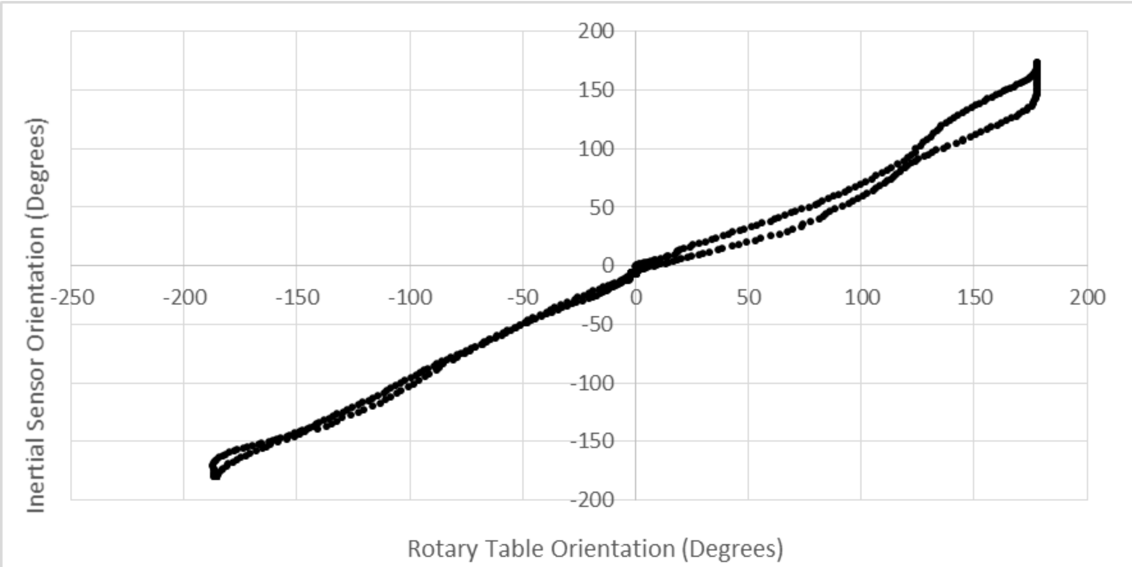

Figure 24 - Rotary table orientation (x-axis) against inertial sensor 1 orientation (y-axis) for heading

Table 9 shows Lin's coefficient of concordance for each sensor and each axis. This test was chosen instead of a correlation test (e.g. Pearson's) as it provides a coefficient relating to how closely two measurements techniques measure the same variable (i.e. 1:1 ratio). This was deemed to be a more statistically robust test than correlation as there can be perfect correlation without having a 1:1 ratio of the measurement techniques. All coefficient of concordance values are greater than 0.99 for every axis indicating that the inertial sensors provide a highly reliable measure of orientation relative to the rotary table. In roll, the mean absolute error ranged from 4.0° for sensor 1 to 5.0° for sensor 2 with a mean absolute percentage error of 1.40%. In pitch, the mean absolute error was calculated to be 3.2° for sensor 4 to 4.0° for sensor 1 with a mean absolute percentage error of 2.2%. In heading, the mean absolute error in this axis was, however, larger than for the other axes. Error in this axis ranged from 12.1° for sensor 3 to 14.3° for sensor 6 with a mean absolute percentage error of 4.0%.

Table 9 - Lin's coefficient of concordance for each sensor and each axis with 95% CIs and mean absolute error in degrees for each sensor and each axis

	Roll		Pitch		Heading	
	Lin's Coefficient of Concordance (95% CI)	**Mean Absolute Error (Degrees)**	**Lin's Coefficient of Concordance (95% CI)**	**Mean Absolute Error (Degrees)**	**Lin's Coefficient of Concordance (95% CI)**	**Mean Absolute Error (Degrees)**
Sensor 1	0.99 (0.99-0.99)	4.0	0.99 (0.99-0.99)	3.0	0.99 (0.99-0.99)	12.9
Sensor 2	0.99 (0.99-0.99)	5.0	0.99 (0.99-0.99)	3.3	0.99 (0.99-0.99)	12.5
Sensor 3	0.99 (0.99-0.99)	4.9	0.99 (0.99-0.99)	3.7	0.99 (0.99-0.99)	12.1
Sensor 4	0.99 (0.99-0.99)	4.4	0.99 (0.99-0.99)	3.2	0.99 (0.99-0.99)	12.2
Sensor 5	0.99 (0.99-0.99)	4.9	0.99 (0.99-0.99)	3.2	0.99 (0.99-0.99)	13.6
Sensor 6	0.99 (0.99-0.99)	4.8	0.99 (0.99-0.99)	3.6	0.99 (0.99-0.99)	14.3

4.3.2. Force-EMG Study Results

4.3.2.1. Subjects

Twelve, healthy male participants were recruited from Cardiff University to take part in the study. Exclusion criteria included a history of any spinal injury, or any indication of neuromusculoskeletal neck problems; for example a limited range of neck motion. Table 10 shows the participant information for the 12 participants that took part in this study. The study was approved by the Cardiff School of Engineering Ethics Committee, with all volunteers providing written consent.

Table 10 - Participant information. Mean data are presented with standard deviations in parentheses

Age (Years)	Height (m)	Weight (kg)	BMI (kgm^{-2})
25.8 (3.59)	1.78 (0.08)	77.9 (11.1)	24.47 (2.33)

4.3.2.2. Force-EMG Correlation Curve

An example of the filtered EMG data and magnitude normalised force data are presented in Figure 25 and Figure 26 respectively. The five force peaks and maximum EMG activity were identified, as can be seen in the figures. The remaining four EMG peaks were identified by using the time variable from the force data. The time delay between each force peak was calculated and the corresponding EMG peaks were identified at the same time intervals, working back from peak muscle activity. This process was conducted for each of the incremental trials, thus yielding 15 force and EMG peaks for each participant. Figure 25 to Figure 31 are for one particular participant. Data for all other participants can be found in the appendices (Figures A.16-A.26).

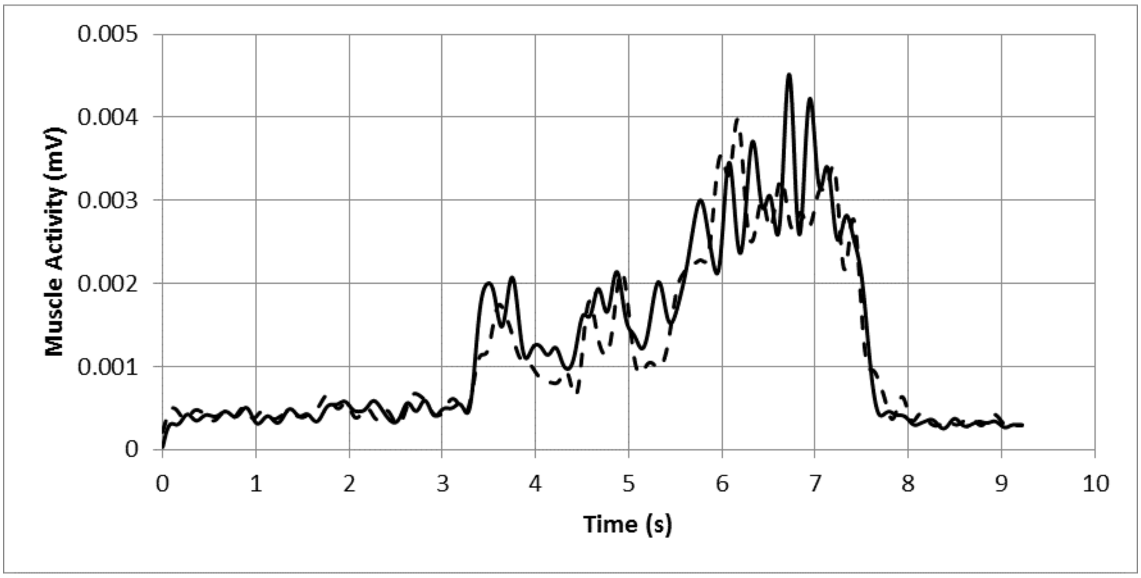

Figure 25 - Filtered right and left CES activity for Participant 2, trial 1. Solid line – Left CES Activity; Dashed line – Right CES Activity

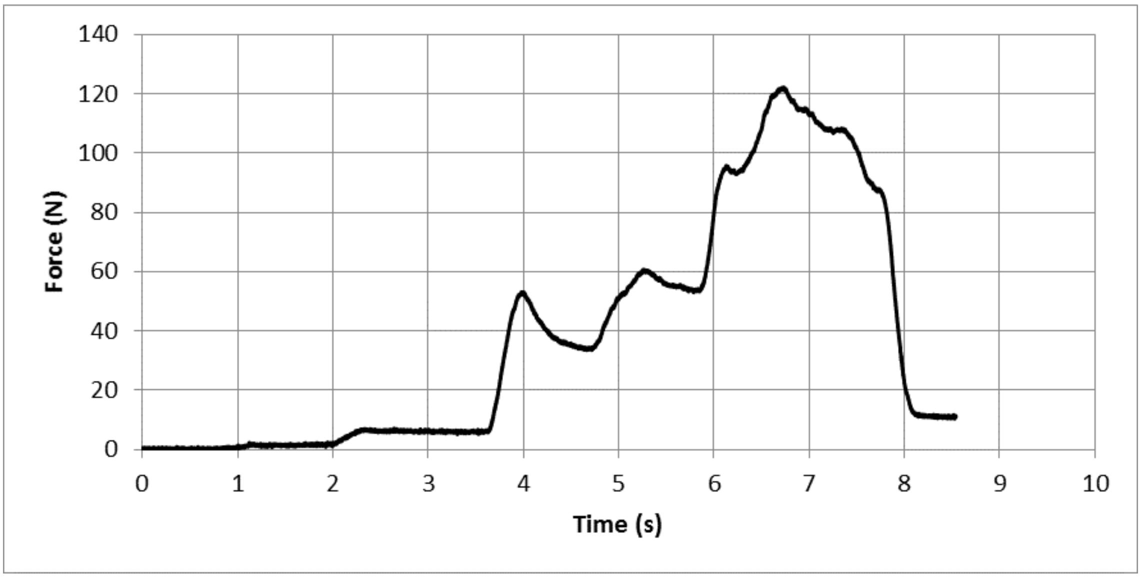

Figure 26 - Vertical force production of the neck extensor muscles for the first trial of Participant 2, trial 1

Having identified the 15 EMG and force peaks, a correlation curve was plotted of the 15 points and a linear line of best fit was superimposed for both left and right cervical erector spinae (CES) muscles for each of the individualised curves using the method of least squares. An example of this is shown in Figure 27. Correlation values for each participant for left (R^2 = 0.76-0.99) and right (R^2 = 0.56-0.97) sides as well as the average (R^2 = 0.67-0.98) of both can be found in

Table 11. The average R^2 value was calculated by averaging the muscle activity of the left and right sides and plotting this against force. The average R^2 value was then calculated from this curve. In excess of 80% of the R^2 values exceeded 0.8, indicating a strong linear relationship between force and EMG. Multiple participants were recruited in recognition of the large variation in EMG amplitude that hinders transferability of correlations between people (Queisser et al. 1994). Although all participants showed a good linear correlation, there was a difference in the slope of the regression lines between participants which may be owing to a number of factors. This approach did, however, introduce inter-participant variation, with, for example, differences in the range of contractions from different persons causing a clustering of data points at lower and higher ends of some EMG-force curves. In an attempt to try and reduce the variability of the data, individualised correlation curves were used to predict force production. Additionally, each participant was instructed to perform isometric contractions up to their MVC, to facilitate identification of a relationship between EMG and force. Adopting this technique of exerting up to MVC does, however, prove difficult to isolate specific muscles for EMG measurements. Thus, participants may have inadvertently recruited surrounding muscles during contractions approaching MVC, to support CES contraction and, thus, aid and increase force production.

The data points for contractions of 50% MVC or less had significantly less deviation from the linear trend line than for contractions greater than 50% MVC. A polynomial model was also fitted to the data during preliminary analysis to determine which model would fit best to the data but the correlation coefficients for the polynomial model were lower than that of the linear. For example, the best polynomial R^2 value was 0.83 whereas the majority of linear R^2 values exceeded 0.8. As a result of this, the linear model was chosen.

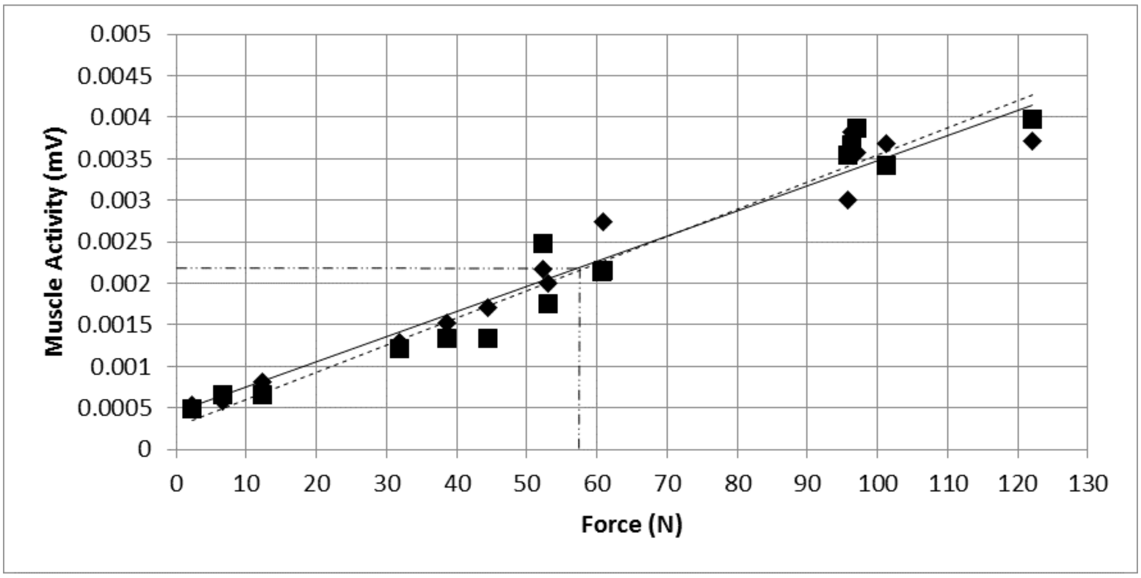

Figure 27 - Linear relationship of force and EMG for all 3 trials for a typical participant. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.95; dotted line – Right side linear trend, R^2=0.96

Table 11 - Correlation Coefficients (R^2) of muscle activity and force for all participants for left and right sides and the average of both

Participant No.	Correlation Coefficient (Left)	Correlation Coefficient (Right)	Correlation Coefficient (Average)
1	0.76	0.82	0.79
2	0.93	0.88	0.91
3	0.88	0.89	0.89
4	0.92	0.93	0.93
5	0.90	0.80	0.85
6	0.86	0.85	0.86
7	0.77	0.56	0.67
8	0.86	0.71	0.79
9	0.63	0.76	0.70
10	0.99	0.97	0.98
11	0.92	0.95	0.94
12	0.72	0.79	0.76

4.3.2.3. Validation of Force-EMG Model

Figure 28 and Figure 29 show the randomised EMG and force traces for one trial of one participant. The randomised contractions were when participants exerted the pre-defined exertions in a random order. The peaks of the EMG and force were easily identified and each EMG peak was used to then predict force using the individualised correlation curve. The researcher was blinded to the force platform data (Figure 30) until a determined value was determined. Having determined a value, the corresponding force peak was identified, assisted by the fact that both sets of data were time normalised to the same number of data points.

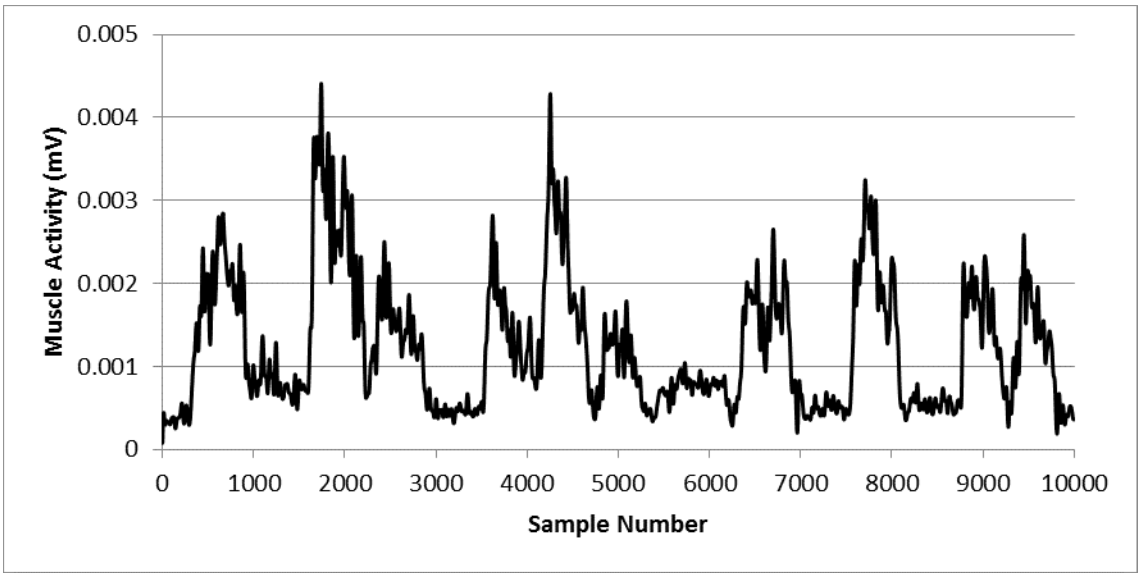

Figure 28 - Right CES Activity for the Randomised Trial of Participant 2

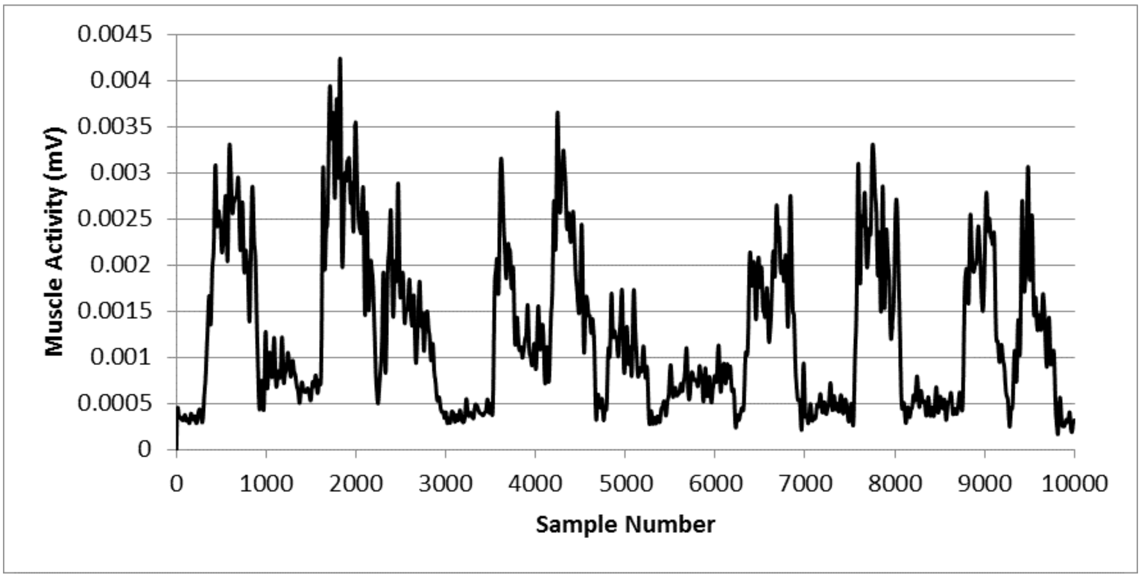

Figure 29 - Left CES Activity for the Randomised Trial of Participant 2

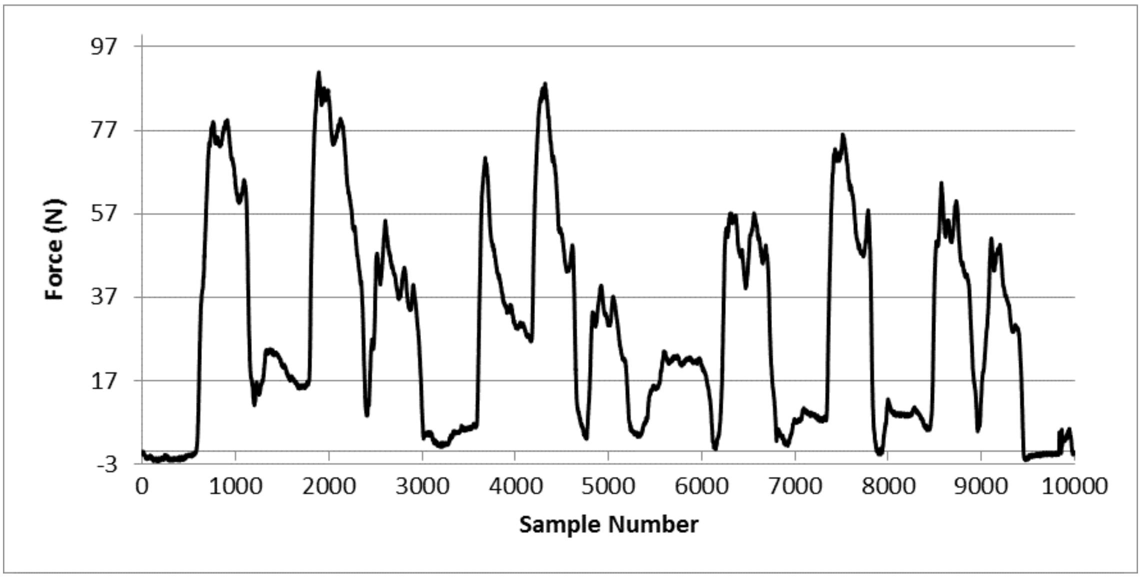

Figure 30 - Vertical Force Production of CES Muscles for Randomised Trial, Participant 2

Determined and actual force measurements for the randomised blinded trials are presented in Figure 31 for this participant. Correlation of determined force and actual force for left, right and an average of both were all extremely high (R^2>0.9).

Table 12 shows the correlation coefficients of all the participants for left, right and an average of both sides. The correlation between determined force and actual force ranged from R^2 = 0.627-0.988 for the left side, and R^2 = 0.563-0.972 for the right side. Combining the sides by averaging resulted in R^2 = 0.730-0.988. There were no significant differences between determined and actual force for all participants (t = 1.598, p = 0.112). The determined force magnitude, however, provided an overestimate of force in most cases. This allowed the calculation of a simple coefficient to adjust the determined force values to obtain greater accuracy. The mean ratio between the determined and actual force values was determined and used as a coefficient to adjust determined values. Data from one participant presented an unrealistic underestimation (outside 2 standard deviations from the mean) and so was not included in the calculation of the mean coefficient. The use of this mean coefficient, to adjust determined force values, resulted in improved force estimation and a much-improved p-value (t = -0.889, p = 0.375). The mean (%) absolute difference for all data points, between actual and adjusted determined muscle force, was 5.80N (18.68%) (95% CI 4.84 – 6.76N; 95% CI 15.98 – 21.39%).

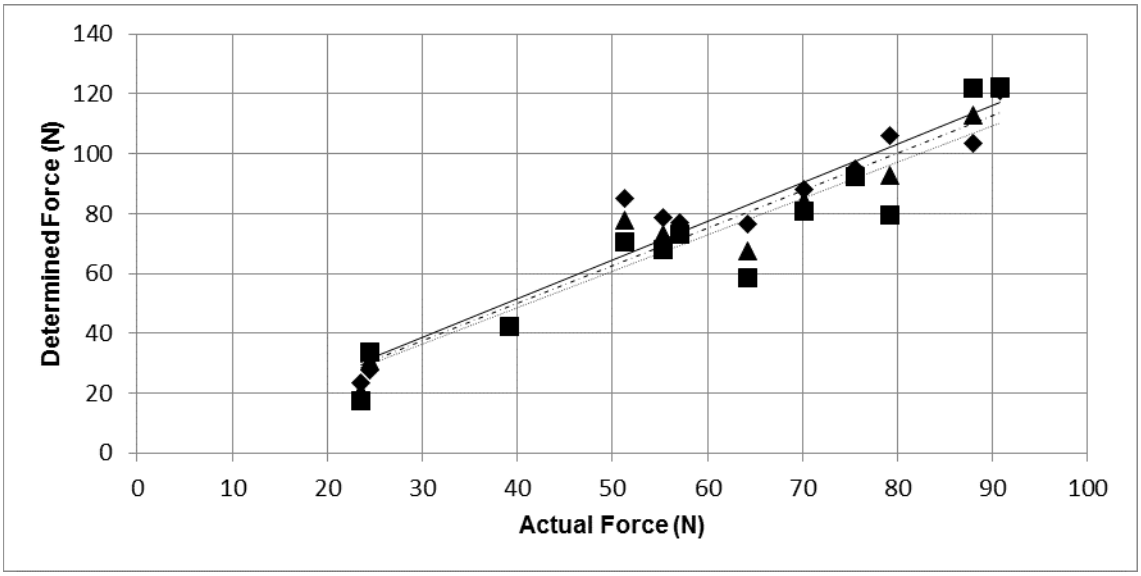

Figure 31 - Determined force against actual force for left and right sides and average of both sides. Diamond – Left side; Square – Right side; Triangle – Average. Solid line – Left side trend line; dotted line – Right side linear trend line; Dash and dotted line – Average linear trend line. (R^2=0.930 (left side); R^2=0.883 (right side); R^2=0.939 (average))

Table 12 - Correlation Coefficients (R^2) of determined force and actual force for all participants for left and right sides and the average of both

Participant No.	Correlation Coefficient (Left)	Correlation Coefficient (Right)	Correlation Coefficient (Average)
1	0.764	0.817	0.862
2	0.930	0.883	0.939
3	0.883	0.892	0.898
4	0.927	0.931	0.944
5	0.902	0.802	0.905
6	0.863	0.853	0.885
7	0.767	0.563	0.730
8	0.856	0.712	0.812
9	0.627	0.762	0.733
10	0.988	0.972	0.988
11	0.916	0.949	0.954
12	0.723	0.786	0.847

A Bland-Altman plot (Bland and Altman 1986) was used to graphically represent the level of agreement between adjusted determined and actual CES muscular force (Figure 32), with the mean difference represented by the solid line. The dashed lines indicate the upper and lower limits of agreement – that is, two standard deviations either side of the mean. Over 90% of data points were within the acceptable limit of error; the upper and lower limits of agreement. It is noticeable, however, that there is much less deviation from the midline for lower contractile forces with scatter become increasingly evident for greater force production.

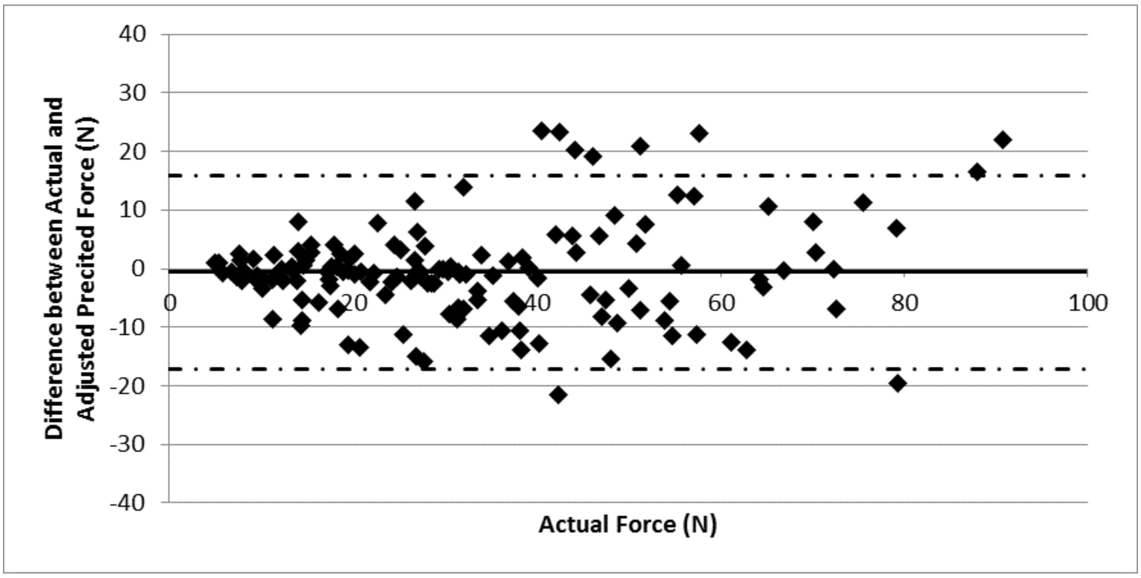

Figure 32 - Bland-Altman plot of differences (actual force vs difference in adjusted determined force) for all 12 participants

4.3.3. Preliminary In-Field Testing

4.3.3.1. Subjects

Nine participants were recruited for this preliminary study. All participants played in the hooker position and had a minimum of two years previous playing experience in the front row. Exclusion criteria included inadequate front row playing experience as per World Rugby guidelines (IRB 2013) which was determined by the team's qualified coach, a history of any major spinal injury, or any indication of current neuromusculoskeletal neck problems; for example a limited range of neck motion. If participants were interested in taking part in the study then an electronic copy of the information sheet was sent a week prior to participation in the study. The participants were given plenty of time to ask any questions arising from reading the information sheet. World Rugby (formerly IRB) laws do not specifically state what inadequate experience is. This is at the discretion of the qualified coach. Table 13 shows the participant information for the nine participants that took part in this preliminary in-field study to demonstrate the general characteristics of the small cohort. The study was approved by the Cardiff School of Engineering Ethics Committee, with all volunteers providing written consent. Details of each individual participant can be found in the appendices (Section A.1.3).

Table 13 - Participant information. Mean data are presented with standard deviations in parentheses

Age (Years)	Height (m)	Weight (kg)	BMI (kgm^{-2})	Playing Experience (years)
20.78 (1.09)	1.82 (0.03)	101.33 (8.65)	30.68 (2.20)	8.69 (4.35)

4.3.3.2. ROM Data

Table 14 provides data relating to the peak normal active ROM for the participants. Mean data are provided with standard deviations in parentheses. Data relating to each individual participant are provided in the appendices. No correlation existed between anthropometric data and peak normal active ROM.

Table 14 - Mean Peak Active Range of Motion of the 9 participants. Mean data (degrees) are presented with standard deviation in parentheses.

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical	46.2 (17.3)	38.5 (14.9)	28.9 (12.8)	26.8 (9.2)	22.3 (12.7)	37.4 (15.4)
Upper Thoracic	30.9 (13.2)	26.9 (6.9)	23.1 (8.4)	26.7 (4.8)	35.1 (17.6)	29.5 (9.1)
Mid-Thoracic	24.1 (11.8)	27.5 (12.1)	22.9 (7.3)	24.8 (12.3)	20.7 (7.7)	21.8 (7.4)
Lower Thoracic	21.2 (13.9)	37.7 (13.4)	31.9 (13.4)	26.7 (6.6)	29.1 (11.2)	29.0 (12.6)
Lumbar	52.4 (13.5)	14.6 (9.4)	29.0 (11.2)	35.2 (18.0)	32.9 (19.4)	31.8 (10.7)

4.3.3.3. Kinematic Data

The kinematic data presented in this section is for one particular participant during machine and live scrummaging for each motion of all spinal segments. Each figure contains a number of different traces and these represent the spinal motion of a specific trial. For example, one figure may show cervical flexion-extension during machine scrummaging for all trials conducted for that participant. The data was time normalised so that the traces could be plotted on the same figure to demonstrate the variation across trials of the same condition with time expressed as a percentage. The reason that time normalisation was necessary was that the length of a scrum was not consistent. Scrums could last from as little as 3-4 seconds to 11-12 seconds and therefore plotting these traces on the same figure would be difficult without time normalisation. During data collection, the data was event marked at the start of the engagement call and again when the ball reached the back of the scrum. For all subsequent figures in the thesis of dynamic ROM, each separate line represents an individual trial. For example, the solid black line in each figure represents data from the same trial so it is possible to determine whether concurrent peaks in ROM are evident across different spinal segments or for different motions of one segment.

For all figures presented in this section, the format is as follows. The top two figures are flexion-extension for machine and live scrummaging respectively, the middle two figures are lateral bending for machine and live scrummaging respectively and the bottom two figures are rotation for machine and live scrummaging respectively. The tables presented show maximum, minimum and the peak-to-peak ROM across the trials presented.

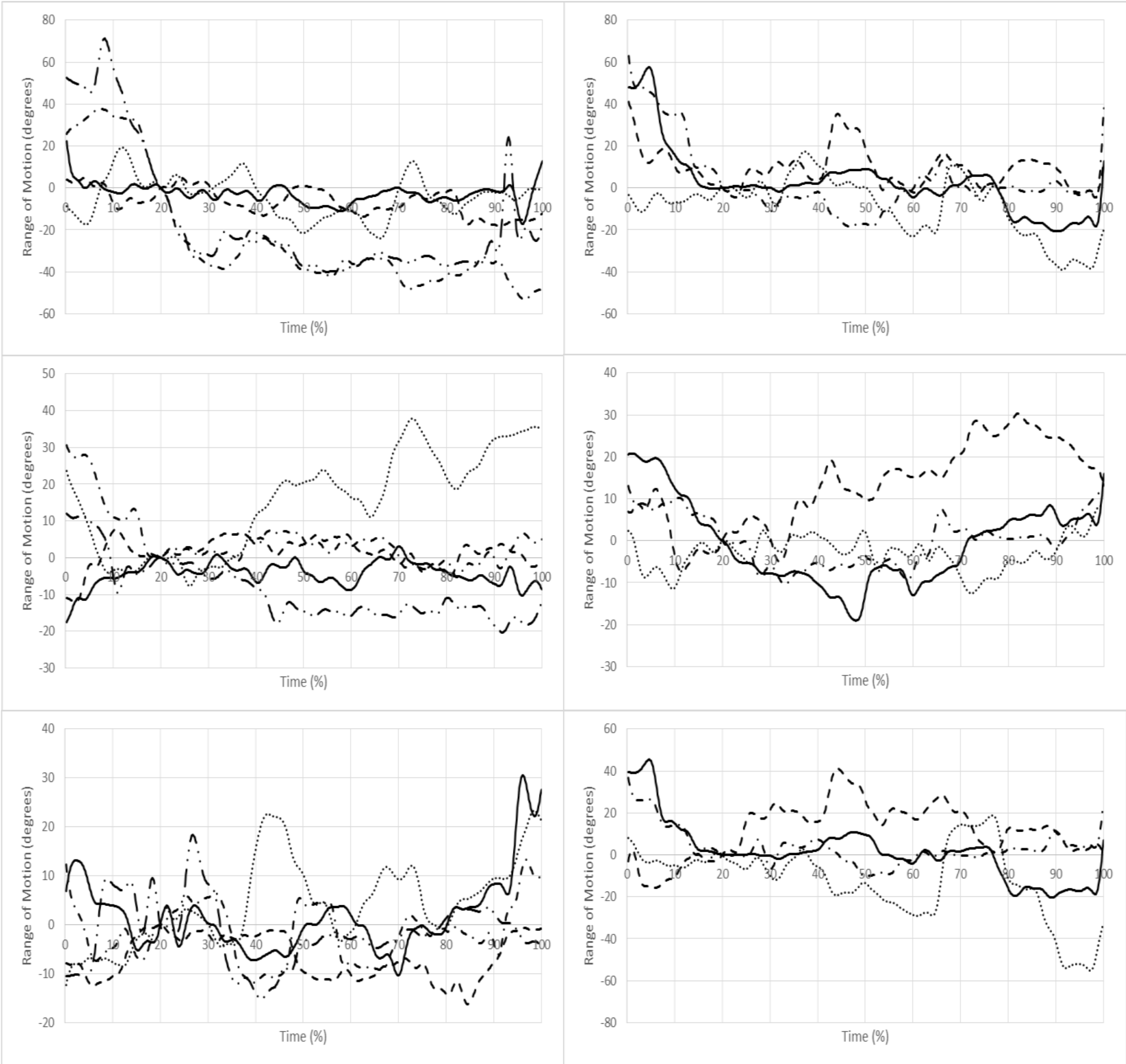

Figure 33 – Dynamic Cervical ROM for Machine and Live Scrummaging

Table 15 - Maximum, Minimum and Peak-to-Peak Cervical ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	Machine	**Live**	**Machine**	**Live**	**Machine**	**Live**
Maximum	70	61	38	30	30	48
Minimum	53	40	20	20	16	52
Peak-to-Peak	123	101	58	50	46	100

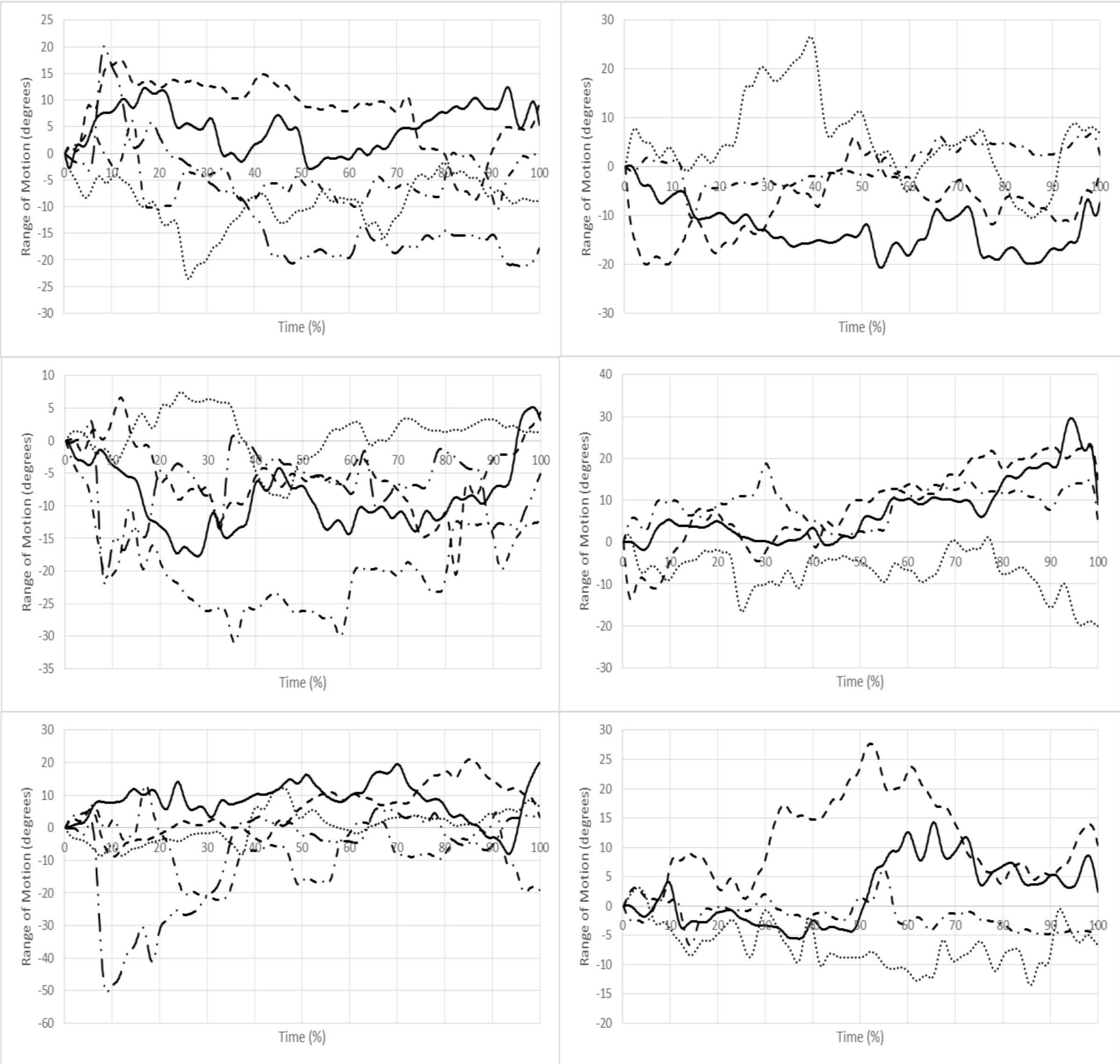

Figure 34 - Dynamic Upper Thoracic ROM for Machine and Live Scrummaging

Table 16 - Maximum, Minimum and Peak-to-Peak Upper Thoracic ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	Machine	**Live**	**Machine**	**Live**	**Machine**	**Live**
Maximum	20	35	31	30	20	25
Minimum	24	20	7	20	50	12
Peak-to-Peak	44	55	38	50	70	37

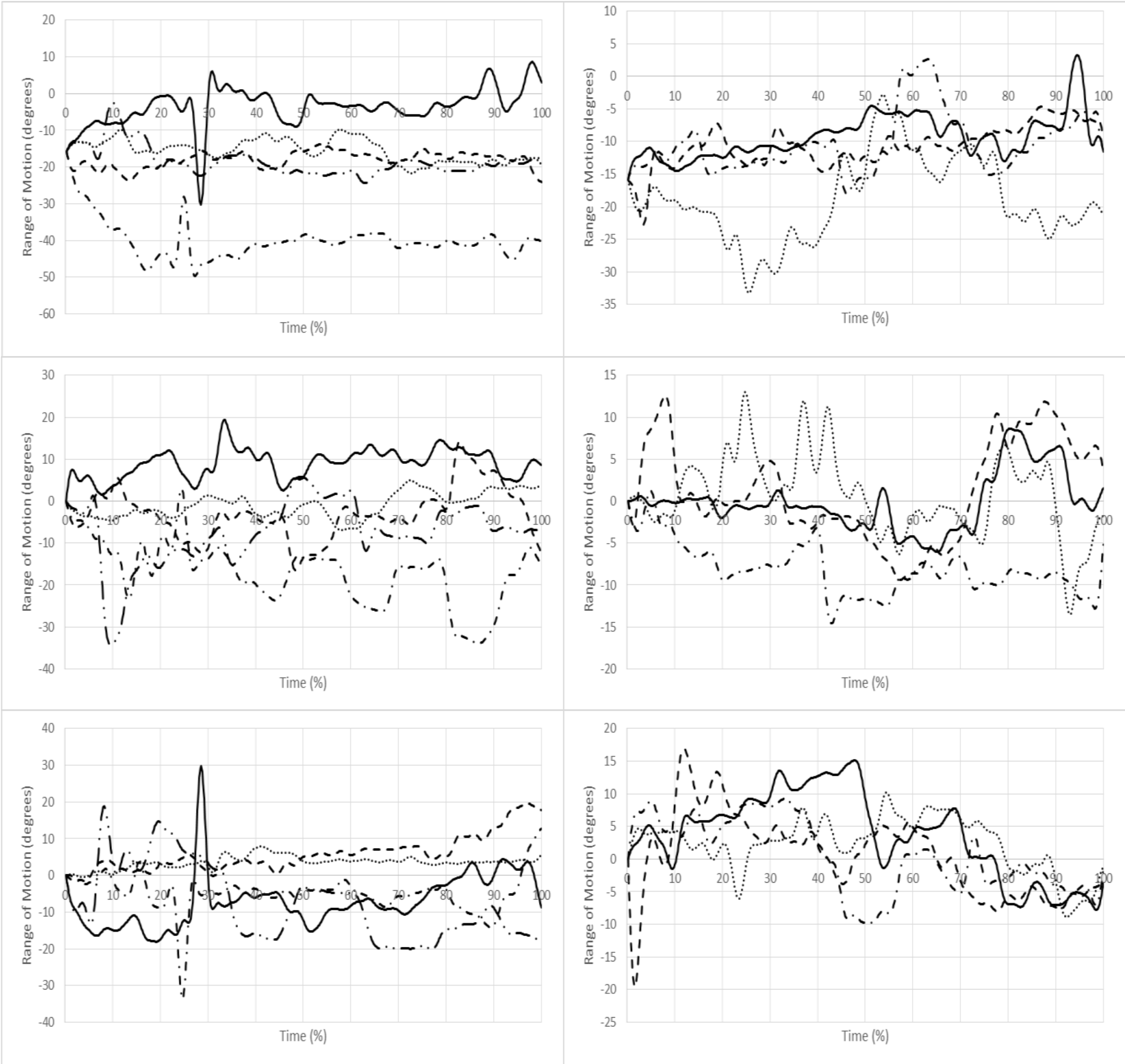

Figure 35 - Dynamic Mid-Thoracic ROM for Machine and Live Scrummaging

Table 17 - Maximum, Minimum and Peak-to-Peak Mid-Thoracic ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	Machine	**Live**	**Machine**	**Live**	**Machine**	**Live**
Maximum	8	4	20	12	30	16
Minimum	50	33	34	15	32	20
Peak-to-Peak	58	37	50	27	62	36

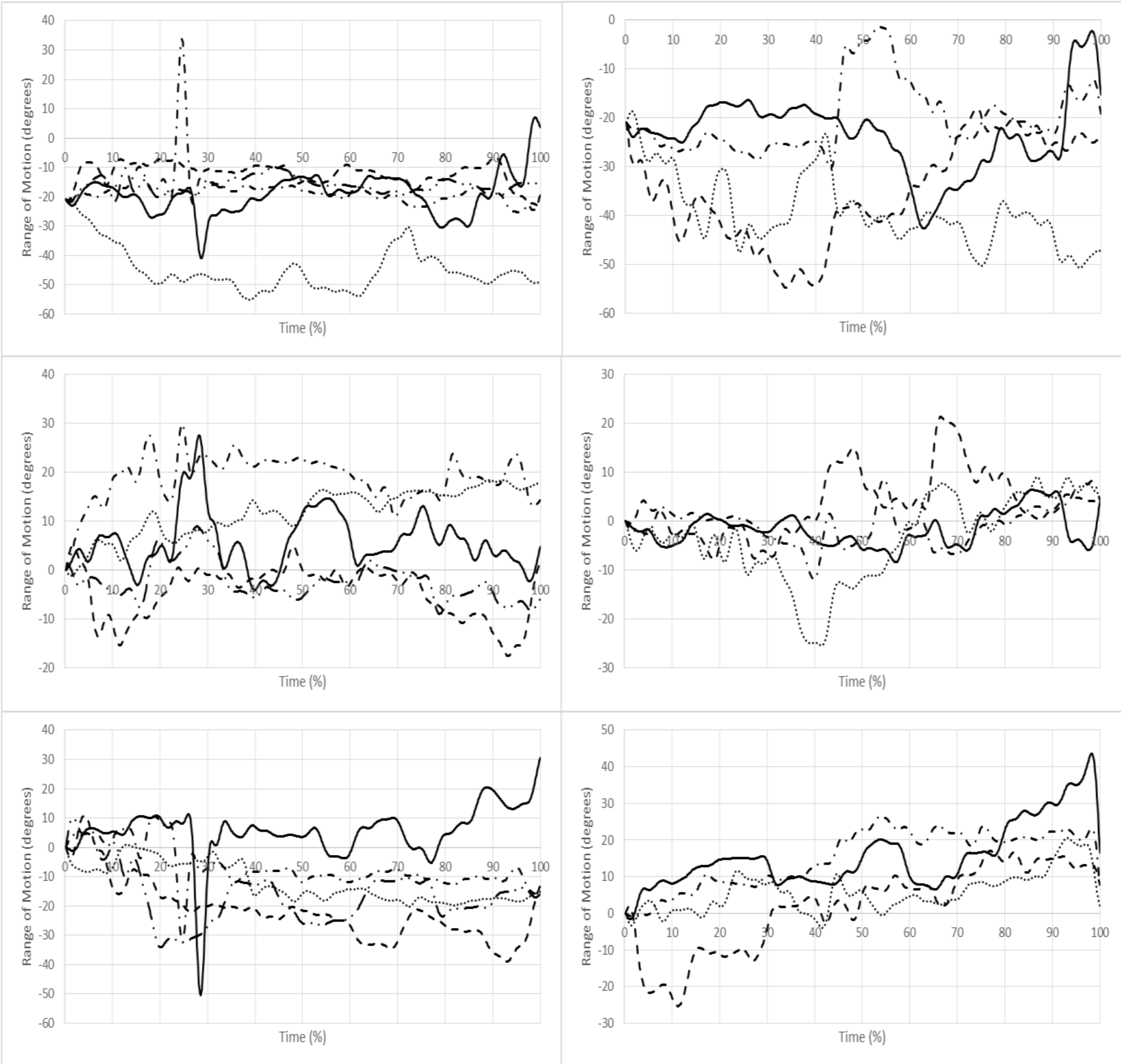

Figure 36 - Dynamic Lower Thoracic ROM for Machine and Live Scrummaging

Table 18 - Maximum, Minimum and Peak-to-Peak Lower Thoracic ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	Machine	**Live**	**Machine**	**Live**	**Machine**	**Live**
Maximum	33	0	30	21	30	43
Minimum	55	54	18	25	50	25
Peak-to-Peak	88	54	48	46	80	68

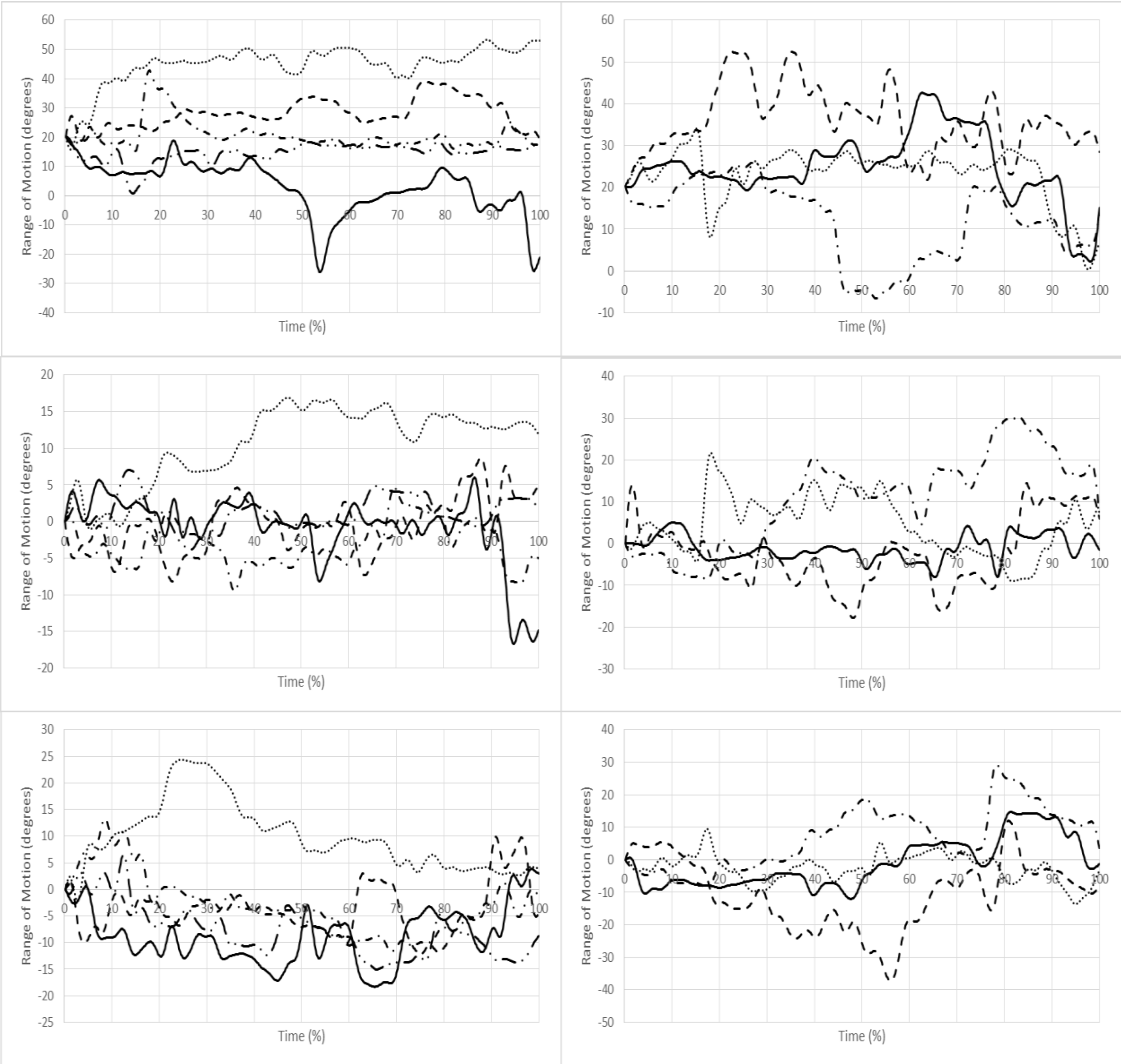

Figure 37 - Dynamic Lumbar ROM for Machine and Live Scrummaging

Table 19 - Maximum, Minimum and Peak-to-Peak Lumbar ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	Machine	**Live**	**Machine**	**Live**	**Machine**	**Live**
Maximum	50	53	17	30	25	28
Minimum	25	4	17	18	18	37
Peak-to-Peak	75	57	34	48	43	65

Peak Kinematic Variables

This section presents data relating to the peak kinematic variables measured for both machine and live scrummaging. These peak variables include peak ROM of each spinal segment in each direction; that is flexion, extension, left and right lateral bending, and left and right rotation. This is in addition to peak rotational velocities that were calculated for both conditions. Peak kinematic data for all participants can be found in the appendices (Tables A.6-A.15).

Table 20 shows mean peak cervical spine kinematics for both machine and live scrummaging. No significant differences (p>0.05) were observed for peak kinematic data.

Table 20 - Mean Peak Cervical Spine Kinematics for both Machine and Live Scrummaging. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		**Angular Velocity (°/s)**	
	Machine	**Live**	**Machine**	**Live**
Flexion	24.8 (7.4)	33.6 (12.4)	13.4 (4.3)	18.5 (3.0)
Extension	39.6 (14.4)	30.4 (14.6)	-14.4 (5.2)	-17.7 (5.2)
Right Lateral Bending	21.5 (8.0)	19.1 (6.3)	7.2 (1.6)	12.0 (5.1)
Left Lateral Bending	17.0 (9.0)	19.1 (8.7)	-6.5 (1.1)	-13.3 (8.1)
Right Rotation	20.3 (10.3)	16.5 (7.5)	12.2 (3.1)	16.3 (3.1)
Left Rotation	26.8 (12.3)	31.0 (11.6)	-10.1 (2.6)	-11.8 (2.1)

Table 21 shows mean peak upper thoracic kinematic data for both machine and live scrummaging. No significant differences (p>0.05) were observed indicating that, for this segment, machine scrummaging accurately represents live scrummaging.

Table 21 - Mean Peak Upper Thoracic Spine Kinematics for both Machine and Live Scrummaging. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	Machine	**Live**	**Machine**	**Live**
Flexion	21.2 (10.4)	15.9 (10.3)	12.2 (6.8)	11.0 (6.6)
Extension	22.8 (13.1)	18.2 (10.7)	-11.0 (6.2)	-10.0 (4.7)
Right Lateral Bending	19.4 (9.8)	14.2 (6.4)	9.0 (4.1)	9.6 (5.8)
Left Lateral Bending	14.8 (7.4)	16.0 (8.3)	-8.5 (3.3)	-9.0 (4.6)
Right Rotation	12.4 (6.6)	17.3 (11.1)	11.9 (5.6)	10.0 (7.2)
Left Rotation	23.5 (14.9)	19.8 (14.0)	-11.7 (5.2)	-10.4 (5.1)

Table *22* shows mean peak mid-thoracic spine kinematics for both machine and live scrummaging. No significant differences were observed ($p>0.05$) for any of the kinematic variables indicating that this segment performs in a similar manner in terms of peak kinematics for both machine and live scrummaging.

Table 22 - Mean Peak Mid-Thoracic Spine Kinematics for both Machine and Live Scrummaging. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	Machine	**Live**	**Machine**	**Live**
Flexion	12.8 (11.8)	10.8 (10.2)	8.8 (2.3)	10.0 (5.2)
Extension	20.0 (11.5)	22.6 (14.7)	-8.0 (2.7)	-9.0 (4.5)
Right Lateral Bending	14.6 (8.2)	13.2 (6.3)	6.3 (2.2)	7.9 (3.0)
Left Lateral Bending	13.9 (7.3)	13.2 (7.5)	-6.7 (2.1)	-9.1 (3.7)
Right Rotation	12.8 (5.3)	10.4 (5.2)	10.5 (4.7)	7.0 (2.5)
Left Rotation	18.2 (5.9)	16.1 (6.1)	-10.4 (4.8)	-7.8 (5.9)

Table 23 shows mean peak lower thoracic kinematic data for both machine and live scrummaging conditions for all 9 participants that participated in this preliminary study. For peak kinematics, no significant differences (p>0.05) were observed for any motion of any variable. This would indicate that this segment responds in a similar manner to both machine and live scrummaging scenarios.

Table 23 - Mean Peak Lower Thoracic Spine Kinematics for both Machine and Live Scrummaging. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	Machine	**Live**	**Machine**	**Live**
Flexion	14.3 (13.9)	10.0 (12.2)	7.8 (2.4)	11.0 (5.8)
Extension	19.8 (15.2)	26.5 (12.9)	-7.9 (3.2)	-8.5 (4.9)
Right Lateral Bending	12.5 (6.8)	14.6 (7.5)	6.0 (2.1)	6.3 (1.9)
Left Lateral Bending	12.0 (1.6)	13.3 (4.7)	-6.0 (1.8)	-8.9 (6.3)
Right Rotation	10.9 (4.2)	15.5 (7.7)	7.2 (2.5)	7.0 (3.2)
Left Rotation	15.8 (7.5)	16.1 (9.2)	-6.8 (3.4)	-7.8 (3.3)

Table 24 shows mean peak kinematic data for the lumbar spine during both machine and live scrummaging. For this segment, the mean peak data does not show any significant differences (p>0.05) for any of the 6 motions or angular velocities. This again would indicate the similar response of this motion segment in terms of kinematics to both machine and live scrummaging.

Table 24 - Mean Peak Lumbar Spine Kinematics for both Machine and Live Scrummaging. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	Machine	**Live**	**Machine**	**Live**
Flexion	36.9 (13.3)	36.8 (6.9)	6.3 (2.6)	8.8 (6.3)
Extension	2.8 (2.9)	3.6 (3.2)	-7.3 (4.2)	-9.3 (5.7)
Right Lateral Bending	14.4 (8.6)	15.3 (7.2)	5.7 (2.4)	8.7 (5.3)
Left Lateral Bending	12.0 (7.0)	14.8 (5.5)	-5.8 (2.9)	-6.8 (2.4)
Right Rotation	15.5 (9.8)	15.5 (7.6)	6.4 (2.9)	9.5 (6.7)
Left Rotation	14.6 (8.1)	12.9 (5.9)	-5.6 (2.5)	-8.2 (6.3)

Kinematic Data Variation

The following tables contain data indicating the variation between trials for each of the 9 participants that took part in this preliminary study. The variation is demonstrated by calculating the coefficient of variation (c_v) which is usually expressed as a percentage. The coefficient of variation is defined as the ratio of the standard deviation (σ) to the mean (μ). That is $c_v = \sigma/\mu$. Some values in the following tables are marked N/A which means that a coefficient of variation value could not be calculated either because only one trial provided data which may be because of problems with sensor adhesion.

Table 25 shows the coefficient of variation values for all 9 participants for each spinal segment during machine and live scrummaging in flexion-extension. For each of the studies, the purpose of presenting coefficient of variation values is to show how variable scrummaging can be between events. The variation ranges from as little as 0.3% for mid-thoracic flexion-extension for participant 2 to 78.7% lower thoracic flexion-extension for participant 8. Both of these extremes were observed for machine scrummaging. When comparing the variation in the data between the two conditions, there were no significant differences ($p>0.05$) between the two conditions. This shows that there is such variation in peak-to-peak flexion-extension ROM between trials and between participants, regardless of condition. This would suggest that all spinal segments respond in different ways from trial to trial and it is very difficult to control this response.

Table 26 shows the coefficient of variation values for all nine participants for each spinal segment during machine and live scrummaging in lateral bending. The variation ranges from as little as 3.3% for participant 4’s cervical segment and as large as 76.7% for participant 4’s mid-thoracic segment. When comparing the variation in the data, there were no significant differences observed ($p>0.05$) between the two conditions when considering variation.

Table 27 shows the coefficient of variation values for all nine participants for each spinal segment during machine and live scrummaging in rotation. The variation ranges from as little as 4.9% for lumbar rotation of participant 5 to as much as 74.5% for upper thoracic rotation of participant 2. There were no significant differences ($p>0.05$) between the two conditions when considering rotational variation of each spinal segment.

Table 25 – Coefficient of Variation Values (%) for all Spinal Segments for Flexion-Extension.

	Cervical		Upper Thoracic		Mid-Thoracic		Lower Thoracic		Lumbar	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	54.5	23.0	37.5	29.5	48.3	22.9	51.7	12.6	34.4	9.4
2	53.2	25.1	20.5	38.3	0.3	18.1	30.2	45.7	17.1	28.2
3	26.3	26.9	6.5	44.4	25.2	55.6	47.2	28.2	56.2	43.9
4	14.8	17.3	14.6	10.0	31.2	14.6	18.8	15.7	22.2	37.8
5	39.5	49.8	44.8	0.9	45.3	N/A	41.0	39.8	50.9	20.9
6	13.3	20.3	49.0	26.0	20.5	38.3	44.5	16.3	40.6	31.6
7	N/A	8.1	20.1	N/A	21.9	42.0	16.7	31.2	43.9	30.7
8	5.2	57.2	24.9	41.9	72.7	44.4	78.7	67.1	73.7	8.1
9	N/A	46.0	48.4	5.8	14.7	5.3	28.2	49.5	32.4	26.3

Table 26 – Coefficient of Variation Values (%) for all Spinal Segments for Lateral Bending.

	Cervical		Upper Thoracic		Mid-Thoracic		Lower Thoracic		Lumbar	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	31.5	27.2	20.0	26.7	39.8	24.3	22.7	30.7	23.9	33.0
2	10.7	66.1	7.9	41.9	24.5	39.4	23.0	48.2	15.3	27.9
3	37.1	46.4	11.9	36.4	12.4	76.7	20.9	39.3	45.8	57.3
4	25.1	3.2	28.1	41.2	17.5	18.0	29.6	26.0	45.8	32.7
5	30.8	36.5	46.4	43.3	21.6	N/A	35.2	9.6	24.5	60.0
6	48.3	29.1	66.5	36.4	50.8	35.0	41.4	41.3	39.2	39.0
7	N/A	23.9	29.0	N/A	30.1	5.5	36.6	45.6	39.8	30.0
8	4.9	53.9	58.8	22.9	71.2	47.0	62.5	45.6	53.6	37.3
9	N/A	7.4	51.5	42.1	38.0	30.9	29.5	36.1	21.2	44.0

Table 27 – Coefficient of Variation Values (%) for all Spinal Segments for Rotation.

	Cervical		Upper Thoracic		Mid-Thoracic		Lower Thoracic		Lumbar	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	28.6	16.5	35.5	28.0	47.3	29.3	40.2	25.9	8.3	29.9
2	6.3	27.8	26.2	38.5	74.5	42.1	43.6	50.3	14.7	45.4
3	23.0	24.3	18.4	21.9	21.8	49.9	16.6	62.2	21.5	33.9
4	43.5	26.5	34.8	54.6	31.0	20.7	18.9	21.0	25.3	46.4
5	37.1	37.0	46.2	46.5	5.5	N/A	37.3	35.0	25.5	4.9
6	27.0	23.9	42.9	26.2	61.6	24.5	69.3	25.9	23.7	54.5
7	N/A	41.0	7.6	N/A	24.7	21.9	14.3	15.5	28.9	31.3
8	7.7	36.6	42.3	16.1	67.2	43.6	15.2	57.9	43.9	28.4
9	N/A	8.0	39.6	14.2	42.3	50.8	14.8	28.6	20.6	38.9

Scrum Collapse Kinematics

During the trials, one scrum collapsed thus providing some extremely valuable data, in terms of the mechanics of potential injury. Although the event is so variable, this particular event provided a unique opportunity to quantify some of the kinematic variables associated with scrum collapse. It is acknowledged that this is an isolated incident and therefore definitive conclusions will be difficult to draw but it is indeed an interesting case and warrants further discussion. Data for all five segments were captured, but it is the cervical and lumbar segments that are of the most interest, because of the history of injuries during this event.

Figure 38 shows dynamic cervical and lumbar ROM during scrum collapse. During this trial, the cervical segment remained, for most of the trial, in slight flexion. When the scrum collapsed, the cervical segment underwent 66° flexion. Furthermore, during this forced flexion the segment was concurrently in 40° right lateral bending and 26° right rotation. The combination of all these motions in an environment where the participant has very little control over their cervical motion has serious injury implications. The 66° of flexion is of particular concern as for this particular participant, their maximum cervical flexion was 70°. This means that the participant was forcibly flexed into a position of over 90% of their normal active cervical range of motion. It should be noted that this participant did not exceed their maximum ROM but, coupled with the driving force of the second and back row players, this situation does present a high risk of injury. The inclusion of lumbar ROM data was to explore whether the lumbar segment collapsed before the cervical segment. In this case, it does not appear that there is any more lumbar flexion at the same time point of cervical flexion. This would indicate that in this particular incident, the lumbar spine does not collapse resulting in the compromised posture of this player. There are many factors that may affect scrum collapse and this kinematic data is just one avenue that has been explored to try and begin to empirically define the cause of scrum collapse. Unfortunately, this data does not help us gain a greater understanding of the causes of scrum collapse but it is, nonetheless, an interesting, albeit, isolated incident.

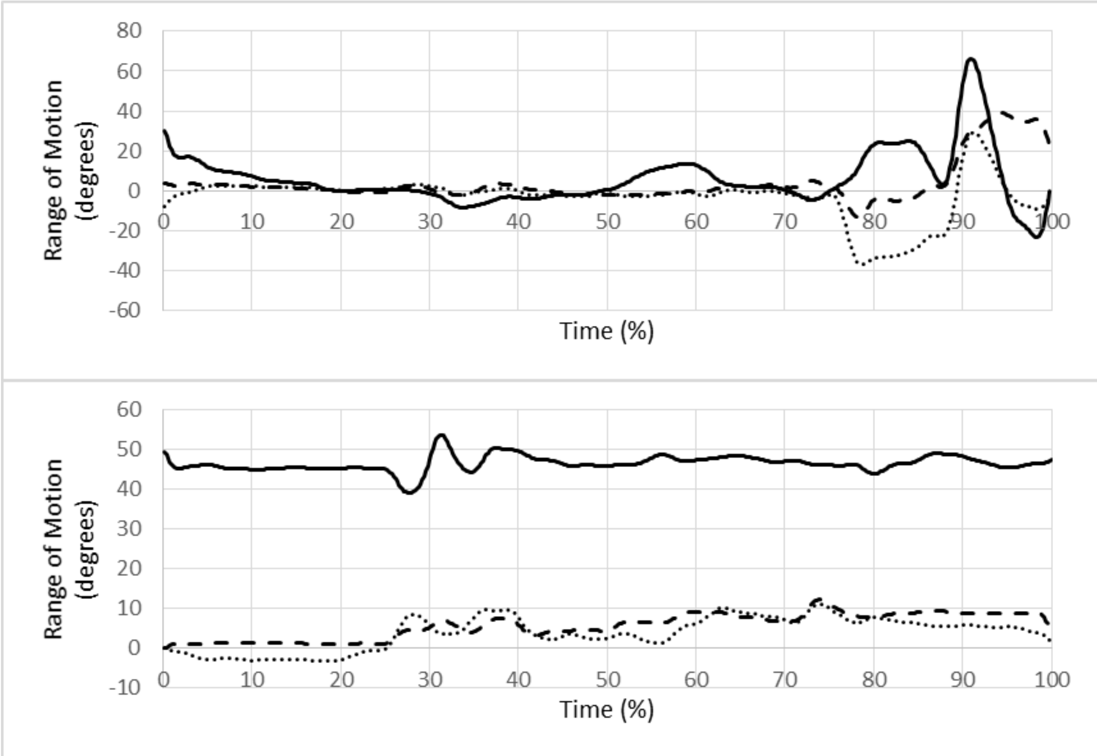

Figure 38 – Dynamic Cervical (top) and Lumbar (bottom) ROM during Scrum Collapse. Solid line – Flexion-Extension; Dashed Line – Lateral Bending; Dotted Line – Rotation

4.3.3.4. Force and EMG Data

This section focusses on the force and EMG data collected during the preliminary in-field trial and the problems that were encountered, particularly with the EMG equipment. Figure 39 shows the incremental force production data for one particular machine scrummaging trial. The peaks in force can be easily identified at five different time points with a maximum vertical force production of 261N.

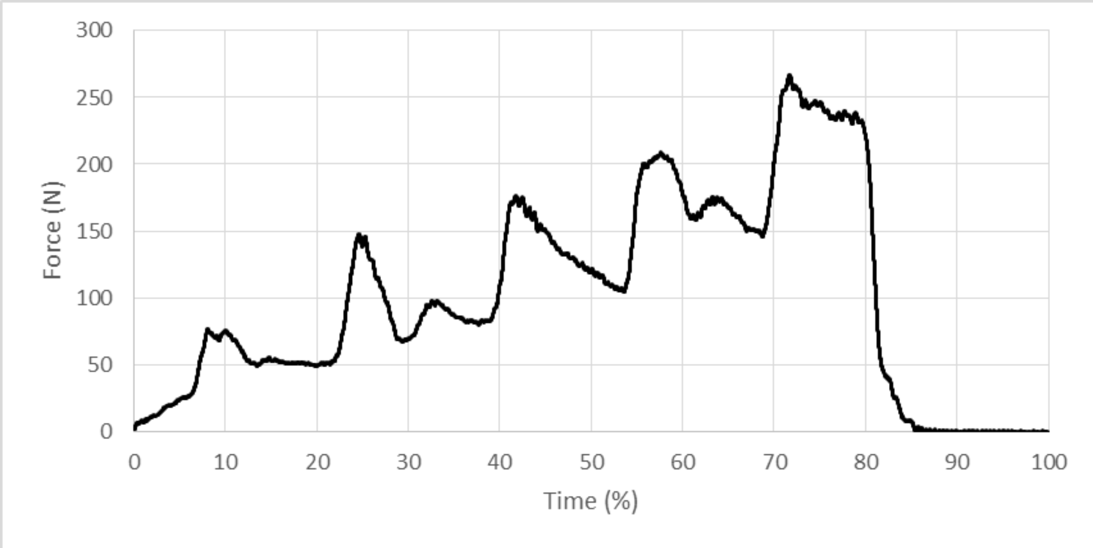

Figure 39 –Example Incremental Vertical Force Production during Machine Scrummaging

Figure 40 shows the incremental increase in EMG activity for both the right (solid line) and left (dotted line) CES muscles during machine scrummaging for the same trial. Distinctive peaks in the data are extremely difficult to identify, and therefore, it was not possible to create an individualised muscle activity and force relationship for use later in the study. This was true for many of the machine scrummaging trials during this preliminary study.

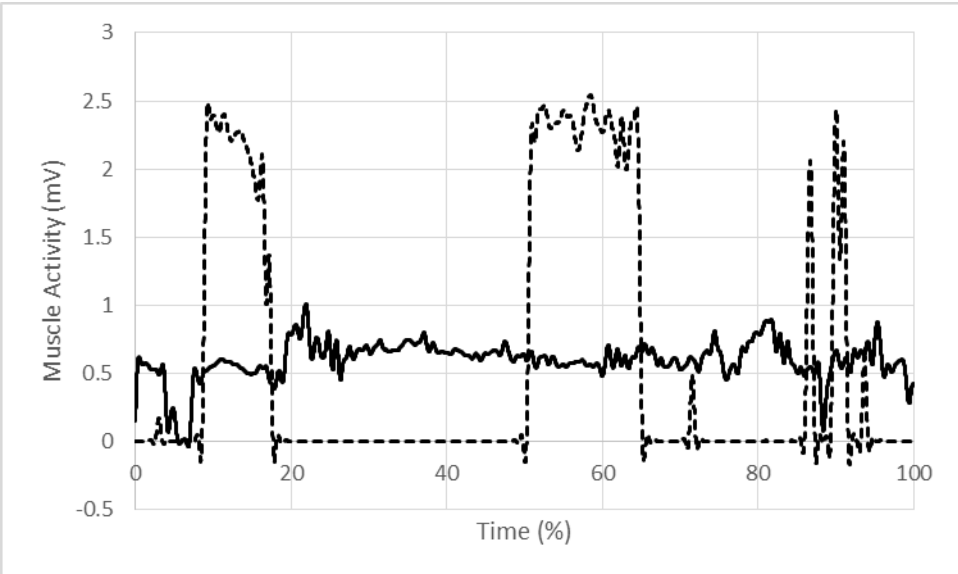

Figure 40 – Example Machine Scrummaging CES Muscle Activity. Solid line - Right CES Activity; Dashed line - Left CES Activity

Furthermore, to demonstrate the poor correlation during these preliminary trials between EMG and force, the variables were plotted against one another (Figure 41). It can be seen from the figure that it is very difficult to identify any sort of coherent correlation between the two variables.

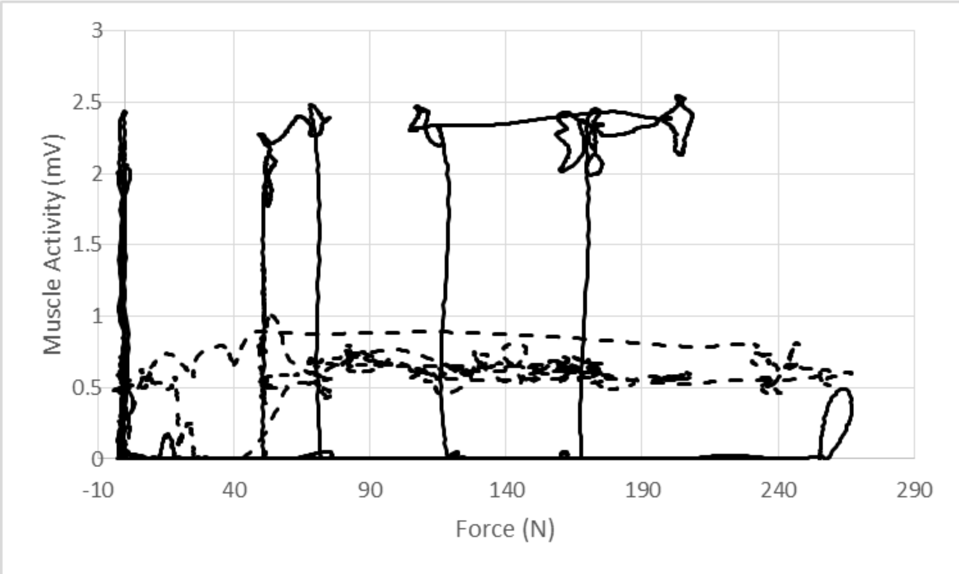

Figure 41 - Force vs EMG showing the poor correlation between the two variables. Solid Line - Right CES Activity; Dashed Line - Left CES Activity.

Despite the poor data obtained for the EMG, the force data was useful and thus, some observations could be made from this. A summary of the vertical force production data for the 9 participants (Table 28) is presented where each participant took part in five machine scrummaging trials. The maximum vertical force production of the CES muscles for the 9 participants ranged from 195.20-471.63N, indicating the large amount of force that can be produced by these muscles. It is likely, however, that the force production was aided by the leg drive of the participants thus, showing an overestimation of neck muscle force production. The CES musculature, however, still must be highly active to resist flexion of the cervical spine against this large vertical force.

Table 28 – Maximum vertical force production of the cervical erector spinae muscles of the 9 Participants during machine scrummaging. Mean data are presented with standard deviations in parentheses.

Participant Number	Vertical Force Production (N)
1	237.40 (18.26)
2	465.18 (118.04)
3	221.96 (28.20)
4	425.56 (39.93)
5	471.63 (69.51)
6	249.42 (35.56)
7	209.20 (21.33)
8	195.20 (14.87)
9	253.40 (28.70)
Mean	303.22 108.72)

4.4. Summary of Results

Absolute errors between orientation measurements of the inertial sensors and the digital encoders of the rotary table ranged between 3.20° (2.20%) for the pitch axis and 14.31° (3.97%) for the heading axis (Table 9). Sources of errors of the inertial sensors are gyroscopic drift and the interference of any ferrous metals. Concordance coefficients for all sensors and each axis were all greater than 0.99.

The mean (%) absolute error between determined and actual force in the force-EMG study was 5.80N (18.68%) (95% CI 4.84 – 6.76N; 95% CI 15.98 – 21.39%). Furthermore, the determined and actual force values were statistically similar ($p>0.05$) and the similarity of these values was strengthened by the use of a simple adjustment coefficient.

The final, preliminary in-field study compared machine and live scrummaging. During this small study, no significant differences ($p>0.05$) in peak kinematic data were identified between the two conditions. Moreover, the coefficient of variation for each motion of each segment did not differ significantly ($p>0.05$) between the two conditions indicating that the way in which participants react to scrummaging is individual and varies from trial to trial. Problems were encountered with the EMG data and thus no comparisons could be made between the two scrummaging conditions. The aim was to obtain similar results as presented in section 4.3.2.3 and use the individual correlation curves to determine vertical force the cervical spine is being subjected to but unfortunately this was not possible.

Lastly, one isolated incident occurred where the scrum collapsed providing a unique insight into the kinematics of this potentially injurious event. The participant in this case was exposed to large magnitudes of coupled motion although their physiological ROM was never exceeded. Combined with the resulting force from the drive of the second and back row players, it may be hypothesised that being exposed to this sort of motion may put the player at a high risk of injury. In vitro studies have shown the strength of the cervical spine to be less resistant to compressive force when in flexion (Przybyla et al. 2007). The biomechanical significance of all these results will be evaluated in the general discussion.

4.5. Conclusions

The primary purpose of this chapter was to develop a method for collecting biomechanical data of the spine during scrummaging. The methods developed in the laboratory provide a useful tool in quantifying spinal biomechanics during rugby scrummaging although a number of problems were encountered in-field. Even though it was not possible to utilise the EMG data, this preliminary study has provided an opportunity to explore the problems encountered and address them for future studies.

There are different techniques that could have been used to investigate spinal biomechanics during scrummaging but many of them are not possible to use because of the constrained environment. The methods developed and the results presented in this chapter provide researchers with a technique to investigate spinal biomechanics, not only in rugby scrummaging, but also in other situations which have similar environmental constraints.

As well as the contribution of the method developed, this chapter also provides valuable data of segmental spinal kinematics during scrummaging. This provides the reader with an insight into spinal kinematics and the individual contributions of spinal segments in a scenario which this has not been previously investigated.

It would have been desirable to determine force production of the CES musculature during scrummaging but numerous problems were encountered which meant that this was not possible. The results did, however, serve as a platform on which the methods could be evaluated to ensure that similar problems were not encountered in future studies.

The next two chapters use these methods to measure and analyse spinal biomechanics of the hooker in a variety of different scenarios.

5. Does Playing Surface Affect Hooker Kinematics

5.1. Introduction

World Rugby, formerly known as the International Rugby Board (IRB), continually reviews the laws to ensure player safety. Scrum stability is an integral part of player safety, as an unstable scrum may expose front row forwards to scenarios that are potentially dangerous (Williams & McKibbin, 1987). Greater vertical and lateral forces may be generated (Milburn and O'Shea 1994) and greater excursion of range of motion (ROM) mean players must make more postural adjustments and are thus, themselves less stable (Cazzola et al. 2015).

World Rugby have also focussed on improving game quality by permitting synthetic turf for use at all playing levels, ensuring a consistent playing surface standard and so encouraging high-quality and faster paced rugby. In the UK, elite teams including Cardiff Blues, London Saracens and Newcastle Falcons have adopted such surfaces now using a '3rd generation' (3G) synthetic turf that comprises a stone base, shock pad, carpet and rubber infill. Such surfaces are specifically designed to more accurately replicate the mechanical response of natural turf (IRB 2003) thereby eradicating the extenuated ball bounce and high injury prevalence associated with earlier generations. The injury prevalence on synthetic turfs is perceived to be higher than on natural turf, however no significant differences have been recorded in the literature (Ekstrand, Timpka, & Hägglund, 2006; Fuller, Clarke, & Molloy, 2010; Steffen, Andersen, & Bahr, 2007; Williams, Trewartha, Kemp, Michell, & Stokes, 2015). Indeed, no researcher has yet been able to identify a definitive cause and effect relationship relating to splay on synthetic turf (Ekstrand et al., 2006; Fuller, Dick, Corlette, & Schmalz, 2007; Steffen, Andersen, & Bahr, 2007).

A high prevalence of injury within the scrum has already prompted other studies to focus on quantifying force production during 'machine-based' scrummaging (Milburn, 1990; Preatoni, Stokes, England, & Trewartha, 2013; Quarrie & Wilson, 2000) and most recently, 'live' scrummaging (Cazzola et al. 2015) environments. The presented study provides a comprehensive analysis of player spinal kinematics during scrummaging on natural versus synthetic turf.

5.2. Aims and Objectives

This study aimed to quantify spinal kinematics during live scrummaging on two different playing surfaces with specific focus on the hooker. Given that there is no significant change in injury incidence between the two types of turf, this study hypothesises that no significant changes will occur in spinal kinematics during scrummaging of the playing position investigated.

5.3. Results

5.3.1. Anthropometric, Background and ROM Data

Table 29 shows the participants' anthropometric and background data. Data includes age, height, weight, BMI, and shoulder, neck and chest circumference. Background data includes playing experiences, training sessions per week and number of scrummages per week. No significant differences were determined between the groups for anthropometric or background data ($p>0.05$).

The mean ROM data for the natural and 3G groups are shown in
Table 30. No significant differences were identified across the peak motion of any segment between the two groups. There was no correlation between peak ROM, anthropometric data and background data suggesting similarity between the groups.

Table 29 – Physical characteristics of the 11 participants from both the 3G and grass trials. Mean data are presented with standard deviation in parentheses.

Group	Age (Years)	Height (m)	Weight (kg)	BMI (kgm^{-2})	Neck Circumference (m)	Shoulder Circumference (m)	Chest Circumference (m)	Playing Experience (Years)	Training Sessions/Week	Scrummages/Week
3G	22.09 (3.78)	1.76 (0.05)	98.00 (13.37)	31.51 (4.38)	0.42 (0.02)	1.25 (0.05)	1.12 (0.06)	7.64 (4.18)	1.73 (0.45)	10.91 (5.38)
Grass	24.73 (4.49)	1.78 (0.04)	99.63 (8.57)	31.52 (2.65)	0.44 (0.02)	1.29 (0.07)	1.15 (0.06)	7.82 (3.93)	1.82 (0.39)	14 .00 (6.47)

Table 30 – Mean Peak Range of Motion of the 11 participants from grass and 3G trials. Mean data (degrees) are presented with standard deviation in parentheses.

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Group	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
Cervical	34.3 (6.4)	37.1 (6.2)	34.1 (10.4)	37.8 (8.2)	40.3 (9.6)	39.8 (7.6)	43.4 (8.3)	41.2 (10.5)	40.1 (6.6)	37.9 (5.2)	31.7 (6.3)	33.8 (5.1)
Upper Thoracic	17.8 (7.7)	17.7 (6.6)	33.4 (11.3)	37.2 (16.7)	21.5 (7.4)	20.5 (6.4)	27.5 (16.0)	24.3 (7.0)	22.5 (8.2)	19.9 (9.8)	26.5 (15.4)	30.0 (9.5)
Lower Thoracic	14.3 (5.0)	22.0 (12.0)	25.4 (10.8)	24.6 (9.0)	15.7 (4.8)	15.9 (1.9)	24.6 (9.8)	20.7 (9.4)	23.6 (10.1)	21.5 (6.9)	24.2 (6.7)	18.6 (3.8)
Upper Lumbar	42.7 (6.6)	47.6 (16.7)	5.2 (1.5)	6.3 (1.1)	16.1 (4.2)	16.2 (2.8)	18.5 (5.1)	15.5 (2.9)	21.3 (6.4)	15.5 (6.9)	17.9 (4.8)	16.9 (2.9)
Lower Lumbar	24.8 (10.0)	21.8 (12.3)	15.9 (7.7)	19.7 (6.7)	18.9 (6.5)	17.1 (3.0)	16.4 (7.1)	18.3 (5.2)	19.9 (9.9)	18.7 (3.9)	25.7 (10.8)	21.6 (3.1)

5.3.2. Kinematic Data

The kinematic data presented in this section is for one particular participant of each group (surface) during live scrummaging for each motion of all spinal segments. Each figure contains a number of different traces and these represent the spinal motion of a specific trial. For example, one figure may show cervical flexion-extension during live scrummaging on grass for all trials conducted for that participant. The data was time normalised so that the traces could be plotted on the same figure to demonstrate the variation. As stated in the previous chapter, this data represents data from a specific trial. That is, the solid black line represents the same trial throughout all dynamic ROM figures.

Qualitatively, it was difficult to identify any specific trends in the dynamic ROM traces, particularly for more mobile spinal segments, which would indicate that each individual scrummaging event is, to some extent, unique. It is therefore difficult to predict specific kinematic traces.

For all figures presented in this section, the format is as follows. The top two figures are flexion-extension for 3G and grass scrummaging respectively, the middle two figures are lateral bending for 3G and grass scrummaging respectively and the bottom two figures are rotation for 3G and grass scrummaging respectively. The tables presented show maximum, minimum and the peak-to-peak ROM across the trials presented.

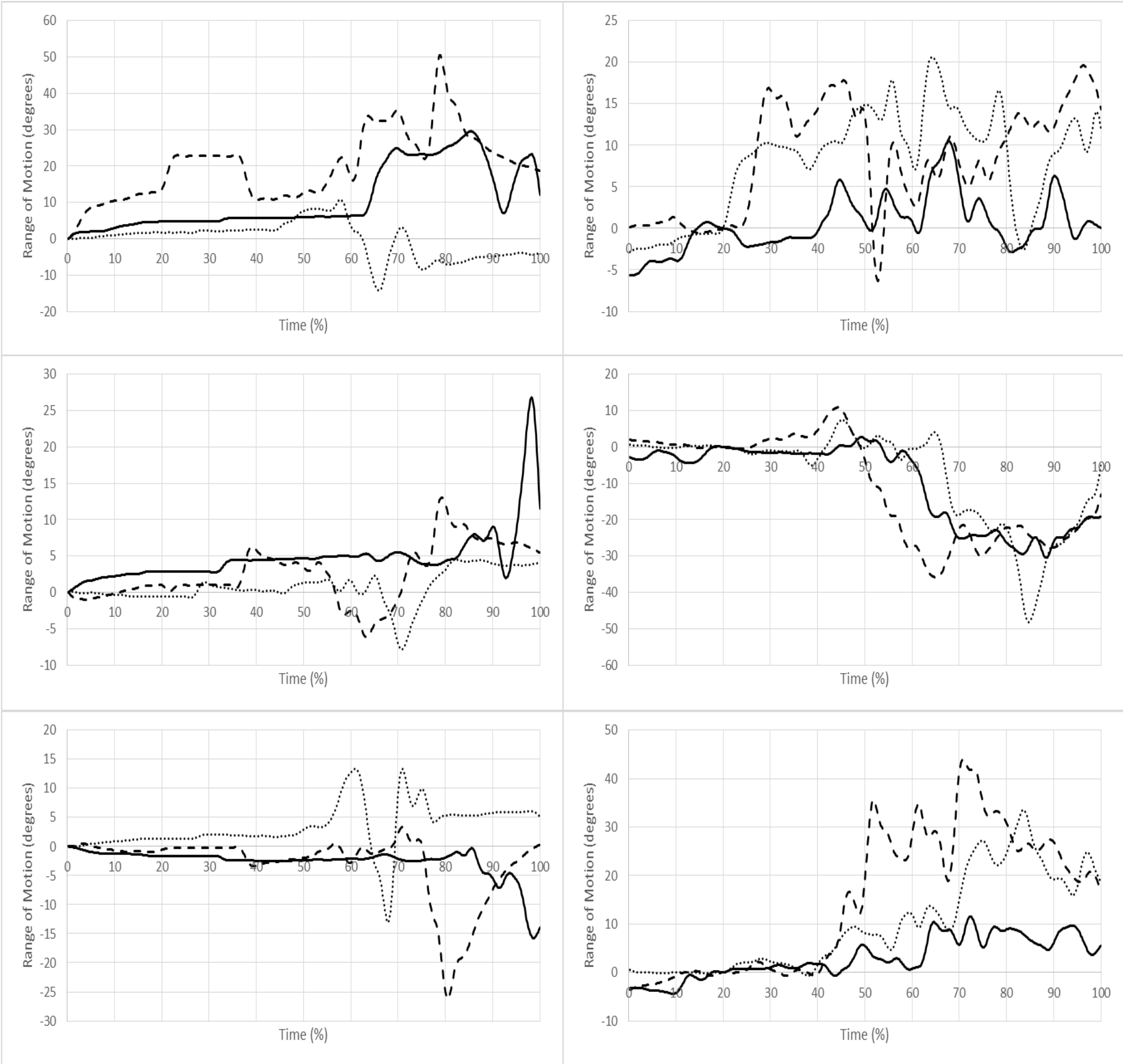

Figure 42 - Dynamic Cervical ROM for 3G and Grass Scrummaging

Table 31 - Maximum, Minimum and Peak-to-Peak Cervical ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**
Maximum	50	20	27	11	13	44
Minimum	14	6	7	48	26	5
Peak-to-Peak	64	26	34	59	39	49

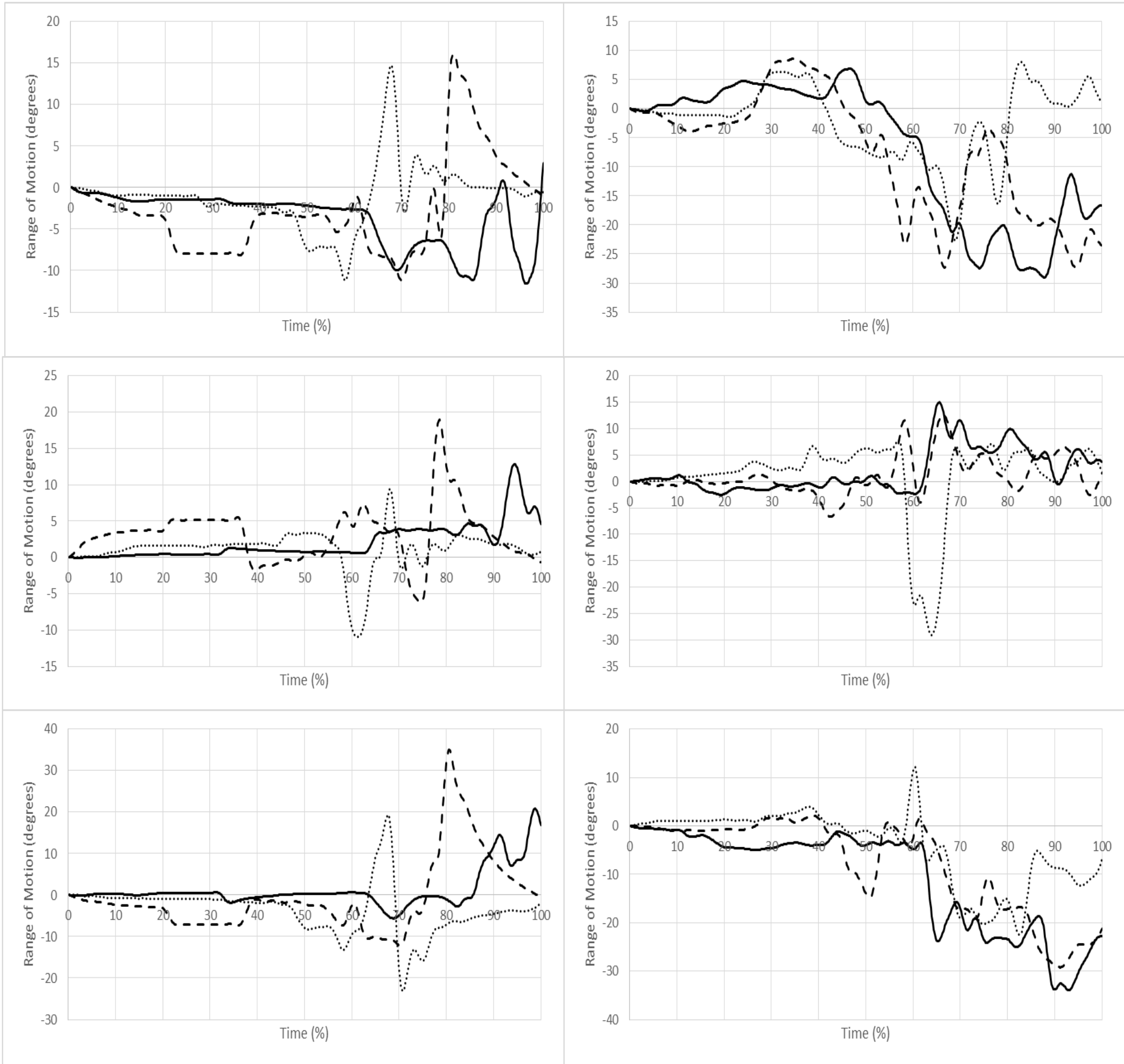

Figure 43 - Dynamic Upper Thoracic ROM for 3G and Grass Scrummaging

Table 32 - Maximum, Minimum and Peak-to-Peak Upper Thoracic ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**
Maximum	15	8	19	15	34	12
Minimum	12	29	11	29	23	34
Peak-to-Peak	27	37	30	44	57	46

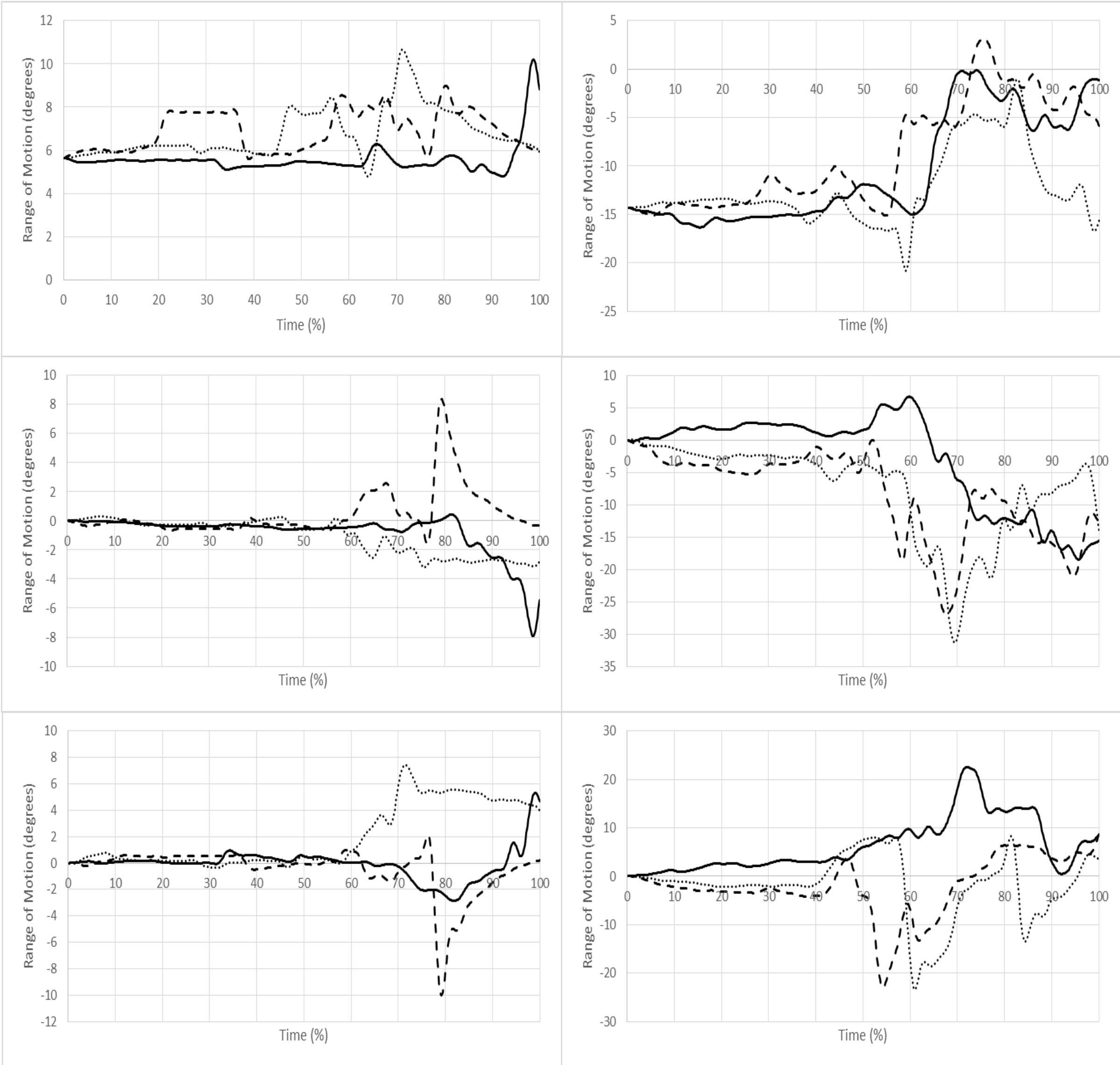

Figure 44 - Dynamic Lower Thoracic ROM for 3G and Grass Scrummaging

Table 33 - Maximum, Minimum and Peak-to-Peak Lower Thoracic ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**
Maximum	11	3	8	7	7	22
Minimum	5 (no –ve)	21	8	31	10	23
Peak-to-Peak	6	24	16	38	17	45

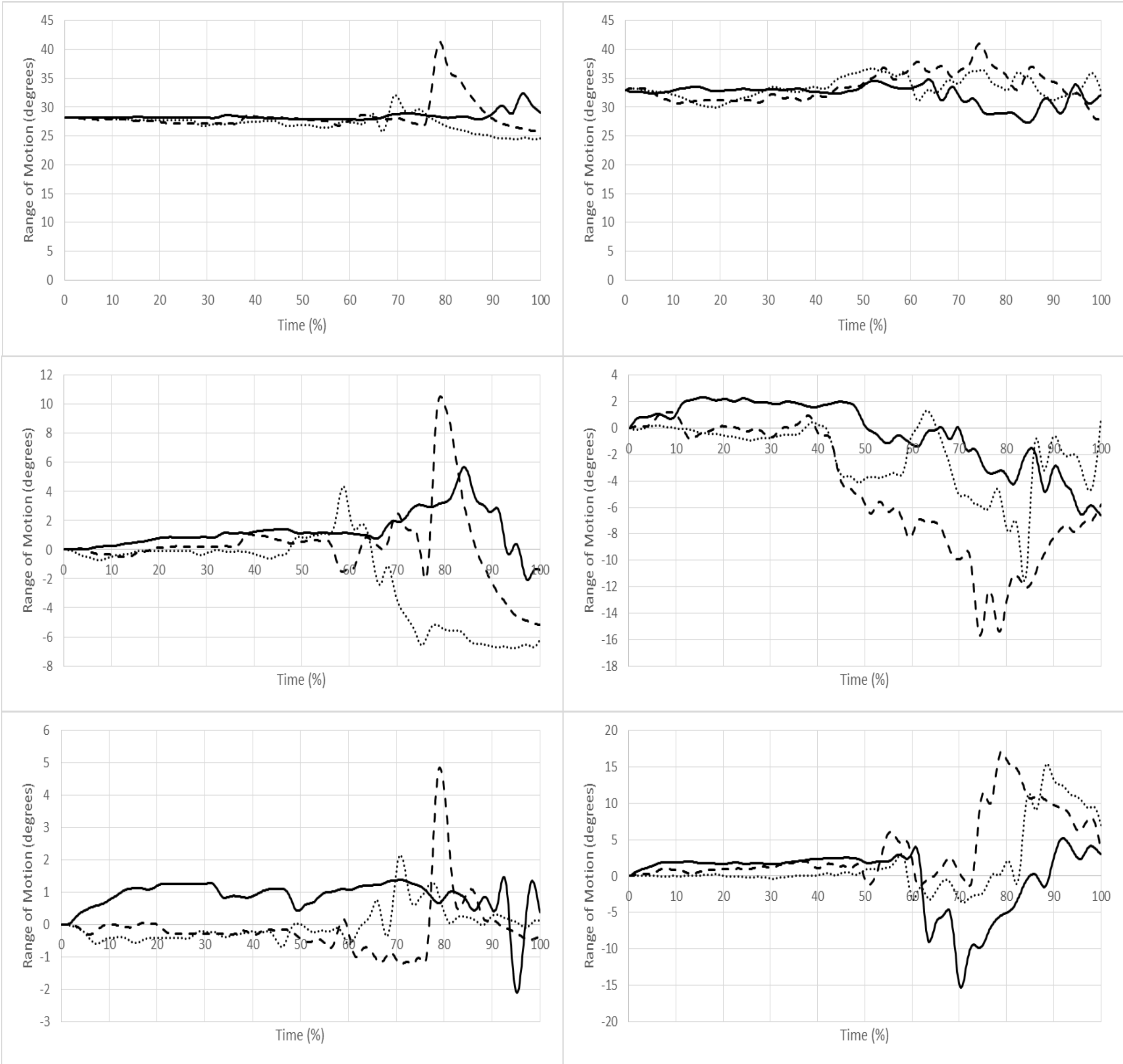

Figure 45 - Dynamic Upper Lumbar ROM for 3G and Grass Scrummaging

Table 34 - Maximum, Minimum and Peak-to-Peak Upper Lumbar ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**
Maximum	41	41	10	2	5	17
Minimum	25 (no – ve)	27 (no – ve)	7	16	2	15
Peak-to-Peak	16	14	17	18	7	32

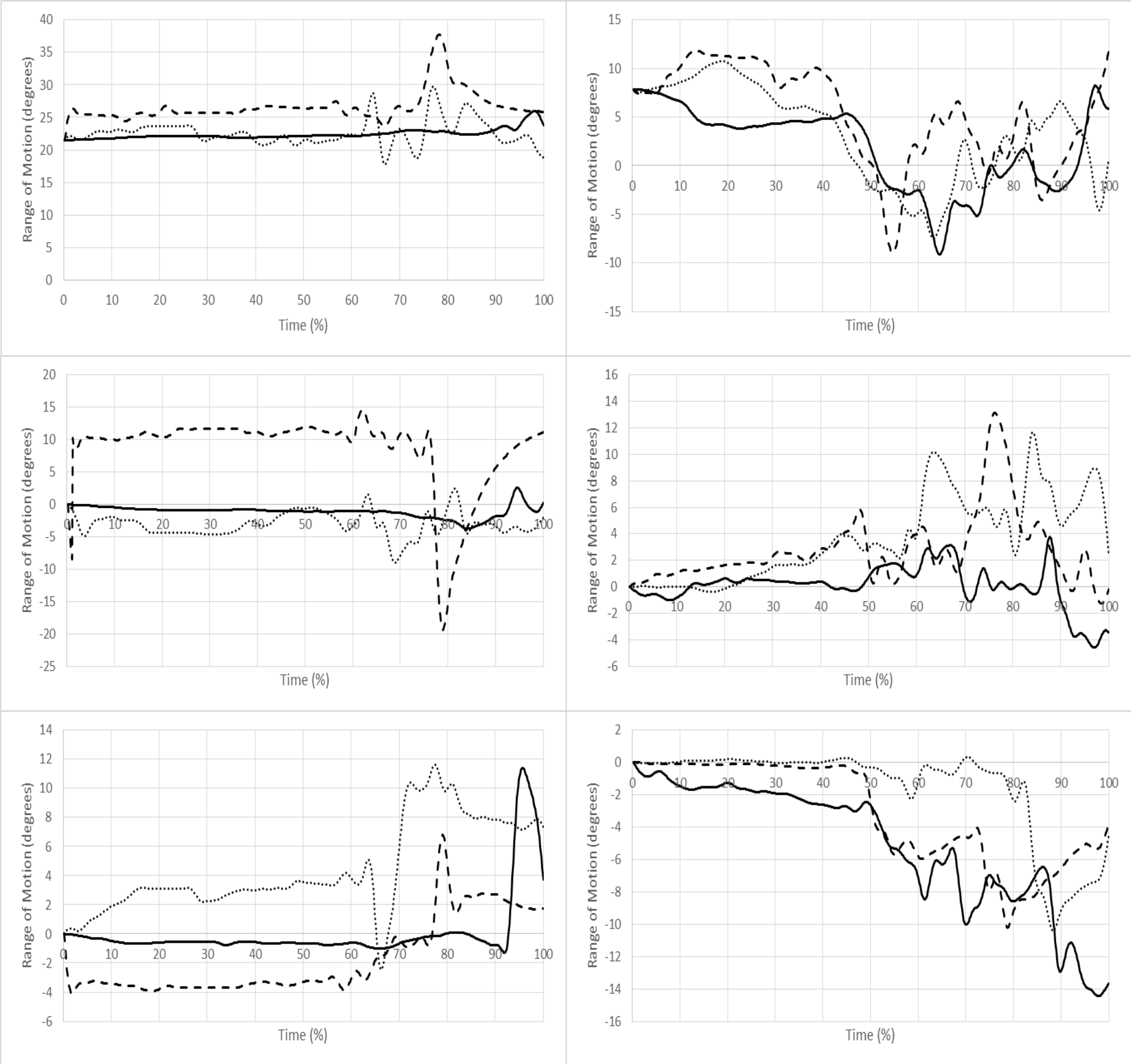

Figure 46 - Dynamic Lower Lumbar ROM for 3G and Grass Scrummaging

Table 35 - Maximum, Minimum and Peak-to-Peak Lower Lumbar ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**
Maximum	37	12	15	13	11	<1
Minimum	17 (no – ve)	9	20	5	4	14
Peak-to-Peak	20	21	35	18	15	15

5.3.3. Peak Kinematic Data

Table 36 shows mean peak cervical spine kinematic data for both 3G and grass surfaces. No significant differences (p>0.01) can be observed for any motion for ROM or angular velocity between the two surfaces. The effect size was also small (d<0.3) when considering peak kinematics variables between conditions.

Table 36 – Mean Peak Cervical Spine Kinematics for both the 3G and Grass Groups. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	3G	**Grass**	**3G**	**Grass**
Flexion	17.08 (9.12)	18.72 (7.54)	8.50 (4.31)	10.26 (4.22)
Extension	19.19 (11.51)	17.95 (10.80)	9.43 (4.15)	10.93 (6.72)
Right Lateral Bending	19.33 (11.67)	19.78 (9.89)	8.16 (3.73)	14.91 (12.96)
Left Lateral Bending	17.78 (13.09)	22.39 (12.15)	7.56 (2.02)	7.13 (1.90)
Right Rotation	19.21 (9.14)	24.45 (7.05)	7.08 (1.88)	10.52 (6.42)
Left Rotation	13.60 (9.71)	11.50 (7.11)	8.15 (4.01)	9.23 (5.09)

Table 37 shows mean peak upper thoracic spine kinematics for both surfaces that were investigated. No significant differences (p>0.01) were observed for any variable for the upper thoracic spine. The effect size was also small (d<0.3) when considering peak kinematics variables between conditions.

Table 37 – Mean Peak Upper Thoracic Spine Kinematics for both the 3G and Grass Groups. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	3G	**Grass**	**3G**	**Grass**
Flexion	8.20 (6.59)	8.80 (7.01)	6.56 (2.18)	10.11 (6.86)
Extension	23.64 (12.38)	21.78 (8.34)	6.87 (3.70)	6.11 (1.41)
Right Lateral Bending	11.33 (6.53)	13.21 (6.43)	5.98 (1.25)	8.03 (4.72)
Left Lateral Bending	13.81 (6.95)	15.99 (12.21)	6.68 (3.78)	9.42 (4.99)
Right Rotation	10.97 (8.75)	14.48 (14.04)	7.55 (5.12)	10.55 (9.66)
Left Rotation	19.45 (7.39)	17.74 (10.26)	7.59 (2.08)	12.04 (12.77)

Table 38 shows mean peak lower thoracic spine kinematics for both 3G and Grass. No significant differences ($p>0.01$) were observed for ROM or angular velocity. The magnitude of the effect, however was large ($d>0.8$) for peak angular velocity of right and left lateral bending and left rotation. The difference in right lateral bending was 3.3°/s, in left lateral bending it was 3.7°/s and in left rotation it was 5.4°/s. All differences were a reduction from grass to 3G surfaces.

Table 38 – Mean Peak Lower Thoracic Spine Kinematics for both the 3G and Grass Groups. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	3G	Grass	3G	Grass
Flexion	8.25 (11.03)	4.59 (5.75)	3.03 (1.56)	4.37 (2.31)
Extension	10.21 (9.19)	15.81 (8.33)	2.77 (1.78)	4.51 (2.83)
Right Lateral Bending	4.97 (3.01)	7.50 (5.32)	3.75 (2.62)*	7.11 (3.53)*
Left Lateral Bending	11.99 (7.63)	16.46 (6.50)	3.31 (1.75)*	6.01 (1.89)*
Right Rotation	10.71 (7.33)	13.26 (12.48)	4.78 (3.52)	7.49 (4.90)
Left Rotation	8.54 (5.42)	13.13 (7.44)	4.36 (2.14)*	9.79 (5.92)*

* - denotes a large effect (d>0.08) between scrummaging on 3G and grass

Table 39 shows mean peak kinematic data for the upper lumbar segment during live (CBS) scrummaging on both surfaces. No significant differences (p>0.01) were observed for any of the kinematic variables that were measured for this segment between the two playing surfaces. The effect size was also small (d<0.3) when considering peak kinematics variables between conditions.

Table 39 – Mean Peak Upper Lumbar Spine Kinematics for both the 3G and Grass Groups. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	3G	Grass	3G	Grass
Flexion	39.62 (13.92)	35.59 (5.52)	4.06 (2.28)	5.37 (2.69)
Extension	0.17 (0.41)	0.52 (0.91)	3.04 (1.64)	4.57 (2.44)
Right Lateral Bending	8.18 (4.45)	9.61 (5.94)	4.10 (1.83)	5.00 (2.20)
Left Lateral Bending	6.85 (4.65)	10.92 (7.03)	3.42 (2.07)	5.89 (4.36)
Right Rotation	9.15 (7.83)	12.38 (8.60)	4.20 (3.19)	5.96 (5.28)
Left Rotation	7.30 (5.18)	8.93 (6.80)	5.03 (4.60)	7.78 (4.33)

Table 40 shows mean peak kinematic data for the lower lumbar segment when performing live (CBS) scrummaging trials on both surfaces. No significant differences (p>0.01) were observed for any variable between the two playing surfaces. The effect size was also small (d<0.3) when considering peak kinematics variables between conditions.

Table 40 – Mean Peak Lower Lumbar Spine Kinematics for both the 3G and Grass Groups. Mean data are presented with standard deviations in parentheses.

	Range of Motion (°)		Angular Velocity (°/s)	
	3G	**Grass**	**3G**	**Grass**
Flexion	10.40 (13.87)	18.34 (15.26)	4.34 (3.45)	5.71 (6.53)
Extension	11.37 (9.33)	9.50 (9.20)	4.24 (2.36)	9.24 (8.73)
Right Lateral Bending	6.69 (4.68)	11.20 (7.77)	3.88 (2.72)	5.97 (5.41)
Left Lateral Bending	8.43 (7.12)	9.15 (8.34)	3.68 (1.90)	5.68 (4.08)
Right Rotation	7.26 (4.58)	10.49 (11.71)	4.83 (4.63)	8.74 (5.47)
Left Rotation	8.92 (7.14)	16.84 (12.41)	3.90 (2.18)	9.19 (7.81)

Table 41 displays mean peak percentage ROM relative to maximum mean peak ROM for all 11 participants of each group on natural and synthetic turf respectively. Peak percentage ROM was calculated from the mean peak ROM experienced during scrummaging for the group and the mean peak ROM of the group demonstrated during the ROM trials. Certain segments, regardless of playing surface utilised a large amount of available ROM. This was particularly evident for the upper lumbar spine in flexion and lateral bending where over 90% of available normal active ROM was utilised.

Table 41 - Mean Peak Percentage ROM during live scrummaging (CBS) for the synthetic (n=11) and natural (n=11) turf trials

	Flexion		Extension		Right Lateral Bending		Left Lateral Bending		Right Rotation		Left Rotation	
	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural
Cervical	47.2	44.5	81.4	64.8	68.7	65.0	65.2	73.6	45.5	54.1	30.2	26.5
Upper Thoracic	46.4	37.1	40.4	46.0	79.7	77.1	70.9	77.3	40.7	62.2	94.6	71.6
Lower Thoracic	50.7	23.8	65.3	36.4	32.1	47.3	74.6	47.5	38.6	39.9	29.5	42.4
Upper Lumbar	88.4	98.0	1.3	3.8	58.9	66.2	55.6	72.3	97.2	95.7	68.1	67.2
Lower Lumbar	42.6	78.5	56.5	42.7	80.8	91.9	75.2	93.6	79.1	20.7	57.1	72.9

5.3.4. Kinematic Data Variation

The following tables contain data indicating the variation between trials for each of the 11 participants that took part in this study. Some values in the following tables are marked N/A which means that a coefficient of variation value could not be calculated either because only one trial provided data which may be because of problems with sensor adhesion.

Table 42 shows the coefficient of variation values for all 11 participants on each surface for each spinal segment during live scrummaging in flexion-extension. The variation ranges from as little as 3.3% for upper thoracic flexion-extension to 72.7% lower lumbar flexion-extension. Both of these extremes were observed for live scrummaging on grass. When comparing the variation in the data between the two playing surfaces, there were no significant differences (p>0.05). This shows that there is such variation in peak-to-peak flexion-extension ROM between trials and between participants, regardless of condition. This would suggest that all spinal segments respond in different ways from trial to trial and it is very difficult to control this response.

Table 43 shows the coefficient of variation values for all nine participants for each spinal segment live scrummaging on 3G and grass in lateral bending. The variation ranges from as little as 3.3% for the upper thoracic segment on grass and as large as 70.7% for the lower lumbar segment on 3G. When comparing the variation in the data, there were no significant differences observed (p>0.05) between the two playing surfaces.

Table 44 shows the coefficient of variation values for each spinal segment during live scrummaging on 3G and grass in rotation. The variation ranges from as little as 0.5% for cervical rotation on grass to as much as 72.9% for lower lumbar rotation on 3G. There were no significant differences (p>0.05) between the two playing surfaces when considering rotational variation of each spinal segment.

Table 42 – Coefficient of Variation Values (%) for all Spinal Segments for Flexion-Extension.

Cervical		Upper Thoracic		Lower Thoracic		Upper Lumbar		Lower Lumbar	
3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**
31.6	N/A	25.3	3.3	22.0	12.7	49.3	9.2	44.4	4.9
4.0	6.6	13.8	14.7	15.2	30.7	12.9	13.0	24.3	13.0
43.4	19.1	28.6	7.2	10.4	8.0	20.0	31.8	30.0	7.7
22.0	39.0	19.2	8.4	28.7	24.1	13.5	9.7	14.8	14.5
11.5	18.5	40.0	35.0	2.9	33.7	13.0	18.3	6.2	20.3
23.1	23.6	16.9	47.9	19.9	15.0	14.4	58.6	27.4	42.6
8.9	22.9	51.9	43.8	8.5	15.7	26.9	55.3	49.3	72.7
28.9	10.1	24.2	23.5	30.1	21.6	10.2	9.6	28.0	27.6
12.8	9.6	17.4	5.5	4.8	29.3	49.6	17.1	23.0	31.0
9.5	25.3	12.4	15.4	3.9	29.3	5.0	13.7	28.2	9.5
14.6	54.2	11.9	32.2	27.4	48.9	13.0	33.6	12.1	5.0

Table 43 – Coefficient of Variation Values (%) for all Spinal Segments for Lateral Bending

Cervical		Upper Thoracic		Lower Thoracic		Upper Lumbar		Lower Lumbar	
3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**
30.2	N/A	26.0	21.3	37.4	38.1	28.4	30.2	69.4	18.5
4.1	7.6	14.9	2.6	10.5	7.0	6.9	3.4	20.8	6.7
15.1	20.5	22.1	35.4	8.4	9.4	20.4	25.2	25.8	31.0
10.4	27.0	4.3	3.3	22.7	8.6	14.4	24.4	23.8	14.9
14.2	27.4	17.6	16.1	39.1	8.9	18.4	16.7	28.2	23.8
28.7	33.9	19.5	26.4	11.1	21.4	63.3	62.2	63.5	35.1
39.9	26.8	52.9	29.8	25.4	41.4	30.7	40.8	25.7	47.5
28.5	18.6	16.1	23.9	34.2	5.9	33.3	26.2	11.3	21.8
11.4	5.8	39.1	22.7	26.9	16.9	64.9	11.9	70.7	20.2
9.8	33.9	12.5	17.9	30.5	49.9	11.0	50.9	21.3	26.4
15.6	16.0	28.4	43.8	27.6	61.5	23.2	49.6	20.9	18.2

Table 44 – Coefficient of Variation Values (%) for all Spinal Segments for Rotation

Cervical		Upper Thoracic		Lower Thoracic		Upper Lumbar		Lower Lumbar	
3G	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**
24.7	N/A	22.9	20.1	20.9	22.3	33.3	16.9	10.4	44.5
29.6	0.5	25.0	14.3	9.4	21.1	72.9	3.9	26.2	1.0
19.8	39.8	6.6	4.3	8.9	15.0	64.5	4.4	12.3	37.7
5.9	24.2	3.4	15.1	25.5	16.8	26.1	20.0	12.4	36.1
25.3	19.9	3.1	24.3	6.4	10.8	41.5	38.9	3.0	26.2
21.9	45.3	16.1	45.3	21.1	60.4	52.2	59.4	49.1	61.4
44.7	31.1	29.4	30.5	39.3	18.1	27.8	57.4	17.5	45.0
24.5	31.9	33.1	36.8	63.0	23.3	37.7	32.3	37.3	44.3
22.2	17.7	10.6	26.5	46.3	15.0	68.9	27.5	40.8	33.0
19.9	21.6	31.7	31.2	34.5	20.8	29.7	18.1	29.4	16.2
11.5	36.1	27.5	40.1	11.0	56.8	29.6	46.6	23.6	8.4

5.4. Summary of Results

When considering scrummaging on two different playing surfaces with regards to hooker spinal kinematics, no significant differences ($p>0.05$) were observed between the two conditions. The synthetic (3G) surface, however, produced more conservative peak ROM values in nearly 80% of the variables measured. The natural (grass) surface produced greater angular velocities for right and left lateral bending and left rotation of the lower thoracic segment. Although these reductions were not statistically significant ($p>0.05$), the size of the effect was large ($d>0.8$).

One key finding was that, regardless of surface, a large proportion of available lumbar spine ROM was utilised. In many cases, over 90% of available ROM was used. The significant of this will be discussed in the general discussion. As with the previous experimental chapter, no significant differences ($p>0.05$) were identified for the variation in data between the two conditions. This would suggest that scrummaging is an event in which individuals react and cope with the constantly changing direction of force in their own way.

5.5. Conclusions

A study was conducted to assess the differences in spinal kinematics of the rugby union hooker when scrummaging on two different playing surfaces; synthetic and natural. A general trend was observed of a reduction in the magnitude of kinematic variables for hookers in both ROM and angular velocity but no significant differences were observed. Only lower thoracic angular velocity, of left and right lateral bending and left rotation proved to have a large effect size ($d>0.8$). These reductions suggest that there is more stability when scrummaging on synthetic pitches compared to natural turf, as players have lower excursions and therefore fewer postural adjustments. It is difficult to stay this with certainty, however, as scrummaging is so variable between specific trials. Given the fact that research suggests there to be no difference in traction on synthetic and natural surfaces, it is suggested that it is the consistency in the properties of synthetic turf in a variety of weather conditions that give rise to this increased spinal stability of the hooker. No significant differences ($p>0.05$) were found between normal ROM values and anthropometric and background data, between the two groups, and no correlation between ROM and anthropometric and background data were observed. Although separate groups were used, the lack of significant differences of anthropometric and background data observed between the two suggest similarity between the groups. It would have been preferable to use the same groups on both turfs and/or a greater number of participants which would have been improvements to the study. It should be noted that the variation on both grass and 3G was large therefore, although kinematic variable magnitude were lower on 3G, the conclusions of this study should be interpreted with caution. Finally, the observed reduction in peak kinematic variable magnitudes suggest that scrummaging on synthetic pitches may potentially be safer, in the long term, as a result of the increased stability that this data suggests although this data should be interpreted with caution..

6. An Investigation Comparing How Different Scrummaging Techniques Affect the Spinal Biomechanics of the Hooker

6.1. Introduction

Rugby is an ever-evolving sport and the laws are constantly reviewed to ensure player safety. Previously, scrum-based research has been conducted on machine scrummaging (Milburn 1993; Du Toit et al. 2005; Quarrie and Wilson 2000) and has focussed on how different factors affect scrummaging power. More recently, however, there has been a significant overhaul of scrummaging laws resulting from a number of research studies with regards to both machine (Preatoni et al. 2015) and live (Cazzola et al. 2015) scrummaging. At the start of the 2013-14 playing season, a new scrum law was introduced worldwide as a result of this research. The engagement sequence was changed from 'crouch-touch-pause-engage' to 'crouch-bind-set'. The 'bind' call means that the props must bind to the shirt of their opposite man, resulting in a reduced distance between the two packs compared to the old laws. It was determined that this change and the requirement of the props to bind, before the engagement, meant that impact force and velocity of engagement was reduced by up to 20% (Cazzola et al. 2015).

Adopting the novel experimental protocol, validated in-field in Chapter 3, spinal biomechanics of the hooker were investigated to determine whether the new regulations are achieving the anticipated biomechanical outcome during live scrummaging. As well as this, spinal biomechanics of the hooker were investigated during machine scrummaging to determine any significant differences in biomechanical loading between machine and live scrummaging

6.2. Results

6.2.1. Anthropometric, Background and ROM Data

Twenty-nine community club and university level hookers were recruited to take part in the study during the 2013/14 playing season and the beginning of the 2014-15 playing season. All participants played in the hooker position as this position. Participants were required to have been suitably trained to play in the front row according to IRB guidelines (IRB 2013). Exclusion criteria for the study included a history of any major spinal injury, or any indication of current neuromusculoskeletal neck problems (e.g. pain).

During the machine scrummaging trials of the 14 participants, EMG and force data was collected where possible. Table 45 summarises the information collected for the machine scrummaging (14 participants) and live scrummaging (29 participants) trials.

Each participant took part in a series of range of motion trials to establish their normal, active spinal range of motion. The data from these trials can be found in Table 46 and Table 47. Table 46 summarises the data for the 14 machine scrummaging participants and Table 47 summarises the data for the 29 live scrummaging participants. No correlation was found between the peak ROM, anthropometric data and playing history. Furthermore, there were no significant differences ($p<0.05$) between the two groups of players. It should be noted that the 14 machine scrummaging participants also all took part in live scrummaging trials and thus, are included in the total of 29 participants in Table 47.

Table 45 – Physical characteristics of the 14 participants from machine scrummaging (1st row) and the live scrummaging (2nd row) trials. Mean data are presented with standard deviation in parentheses

	Age (Years)	Height (m)	Weight (kg)	BMI (kgm^{-2})	Neck Circumference (m)	Shoulder Circumference (m)	Chest Circumference (m)	Playing Experience (Years)	Training Sessions/Week	Scrummages/Week
Machine Group	23.43 (4.17)	1.76 (0.04)	101.09 (12.80)	32.56 (4.04)	0.43 (0.02)	1.28 (0.08)	1.13 (0.07)	7.07 (4.08)	1.64 (0.48)	11.07 (5.08)
Live Group	23.98 (4.06)	1.77 (0.04)	99.33 (11.22)	31.57 (3.55)	0.44 (0.02)	1.28 (0.07)	1.14 (0.07)	7.72 (3.65)	1.83 (0.38)	13.59 (6.10)

Table 46 – Mean Peak Range of Motion of the 14 participants that took part in all 3 scrummaging conditions. Mean data (degrees) are presented with standard deviation in parentheses

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	38.41 (7.85)	24.09 (10.23)	29.18 (10.84)	27.14 (8.77)	44.80 (10.37)	44.29 (10.00)
Cervical (Standing)	43.65 (8.73)	33.79 (8.10)	34.51 (5.38)	30.47 (6.07)	45.47 (9.66)	43.07 (7.64)
Upper Thoracic	16.72 (11.72)	8.63 (11.33)	8.71 (3.26)	6.44 (2.11)	20.44 (11.46)	18.35 (7.72)
Lower Thoracic	16.19 (8.43)	6.74 (4.67)	15.88 (8.44)	16.98 (6.49)	31.31. (11.28)	32.63 (11.99)
Upper Lumbar	33.99 (11.98)	13.98 (7.75)	13.49 (3.98)	14.33 (4.92)	9.65 (5.70)	11.53 (6.55)
Lower Lumbar	26.89 (11.27)	22.08 (13.44)	7.55 (4.31)	7.78 (4.50)	9.69 (4.27)	9.02 (3.81)

Table 47 – Mean Peak Range of Motion of the 29 participants that took part in the live scrummaging trials. Mean data (degrees) are presented with standard deviation in parentheses

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	38.30 (7.45)	26.00 (9.36)	29.30 (8.92)	28.84 (7.76)	42.36 (8.62)	42.78 (8.79)
Cervical (Standing)	43.37 (9.48)	34.73 (10.17)	34.92 (4.63)	32.21 (6.17)	41.42 (8.73)	41.54 (7.58)
Upper Thoracic	18.93 (11.26)	4.83 (8.76)	10.15 (3.51)	9.41 (4.30)	24.95 (11.49)	23.70 (10.20)
Lower Thoracic	17.40 (8..26)	6.47 (5.28)	14.71 (6.48)	15.24 (5.35)	27.32 (12.26)	27.35 (11.84)
Upper Lumbar	31.72 (9.40)	12.06 (7.53)	13.89 (3.93)	13.30 (5.55)	11.05 (6.87)	11.17 (6.56)
Lower Lumbar	26.61 (11.62)	20.70 (10.69)	8.42 (4.14)	8.08 (3.95)	9.95 (4.26)	10.26 (4.30)

6.2.2. Kinematic Data

The kinematic data presented in this section is for one particular participant for each condition investigated. This includes machine scrummaging and live scrummaging of old (CTPE) and new (CBS) engagement sequences. Each figure contains a number of different traces and these represent the spinal motion of a specific trial. For example, one figure may show cervical flexion-extension during machine scrummaging for all trials conducted for that participant. The data was time normalised so that the traces could be plotted on the same figure to demonstrate the variation. Machine scrummaging data is presented here as it was used as part of the study used to determine a calibration curve for participants where possible. Furthermore, this machine scrummaging data is different to that which has been previously presented as it was done using the CBS engagement sequence as opposed to the CTPE engagement sequence in the preliminary study

6.2.2.1. Machine Scrummaging

For all figures presented in this section, the format is as follows. The top figure is flexion-extension, the middle figure is lateral bending respectively and the bottom figure is rotation for machine scrummaging. The tables presented show maximum, minimum and the peak-to-peak ROM across the trials presented.

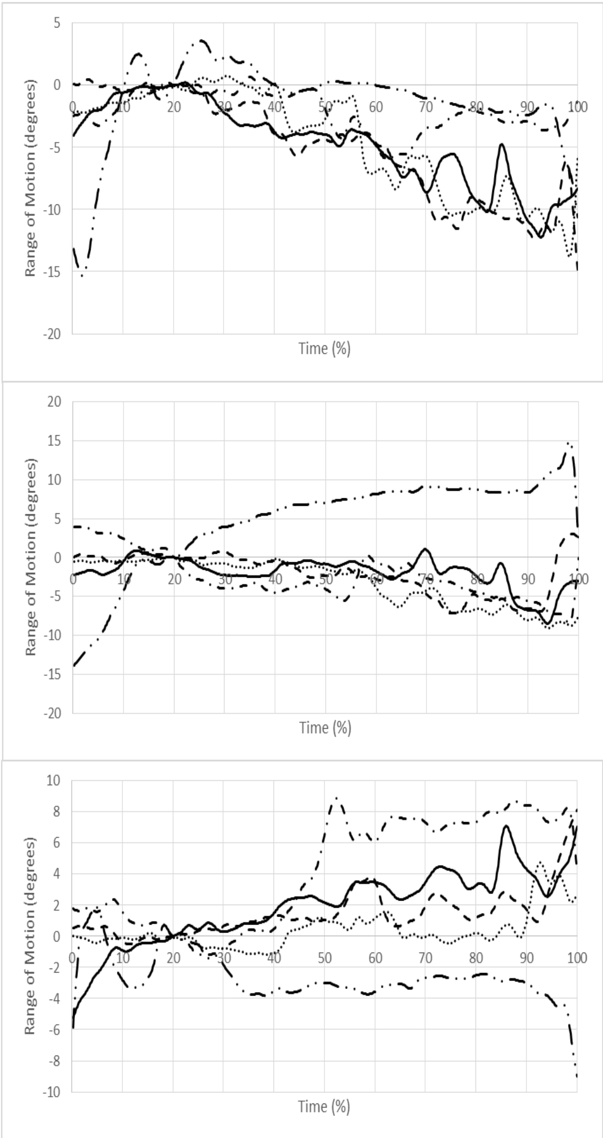

Figure 47 - Dynamic Cervical ROM for Machine Scrummaging

Table 48 - Maximum, Minimum and Peak-to-Peak Cervical ROM

	Flexion-Extension (°)	Lateral Bending (°)	Rotation (°)
Maximum	4	15	9
Minimum	16	15	9
Peak-to-Peak	20	30	18

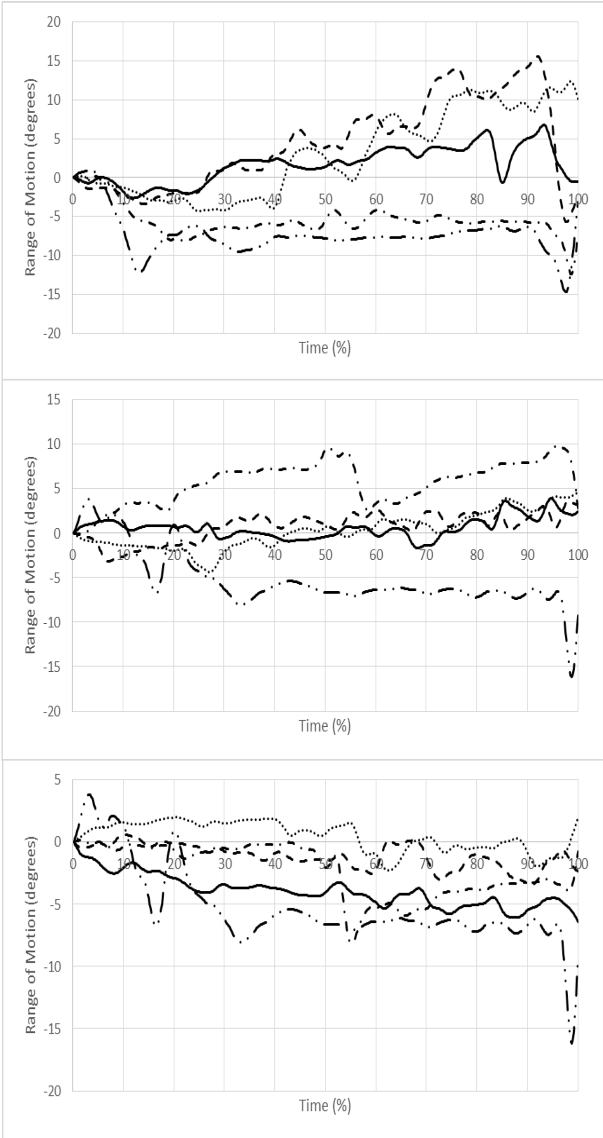

Figure 48 - Dynamic Upper Thoracic ROM for Machine Scrummaging

Table 49 - Maximum, Minimum and Peak-to-Peak Upper Thoracic ROM

	Flexion-Extension (°)	Lateral Bending (°)	Rotation (°)
Maximum	15	10	4
Minimum	15	16	16
Peak-to-Peak	30	26	20

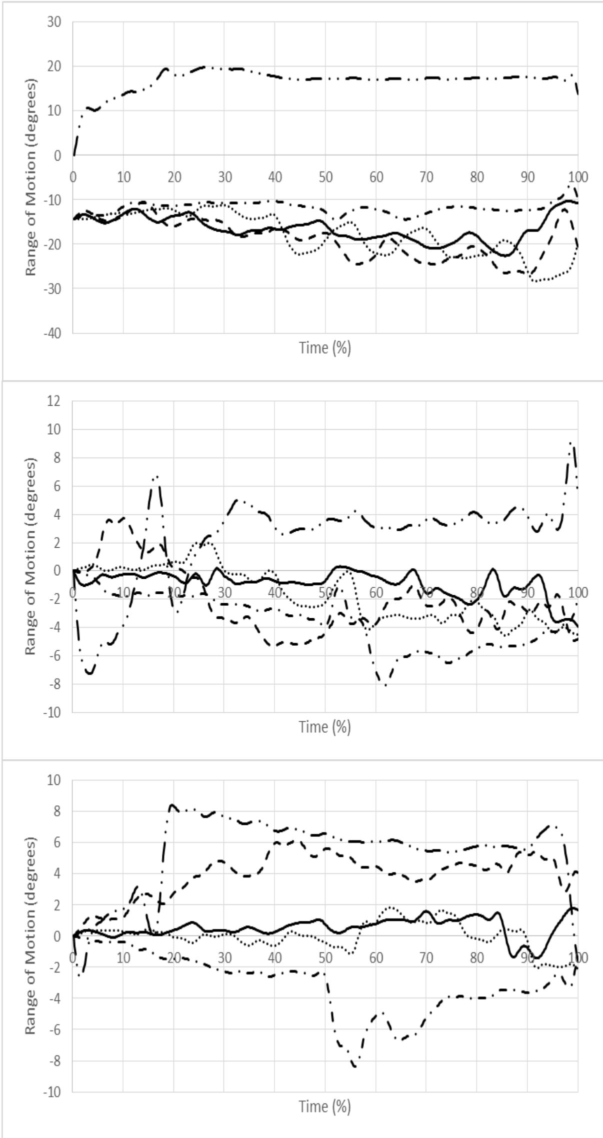

Figure 49 - Dynamic Lower Thoracic ROM for Machine Scrummaging

Table 50 - Maximum, Minimum and Peak-to-Peak Lower Thoracic ROM

	Flexion-Extension (°)	Lateral Bending (°)	Rotation (°)
Maximum	20	9	8
Minimum	26	8	9
Peak-to-Peak	46	17	17

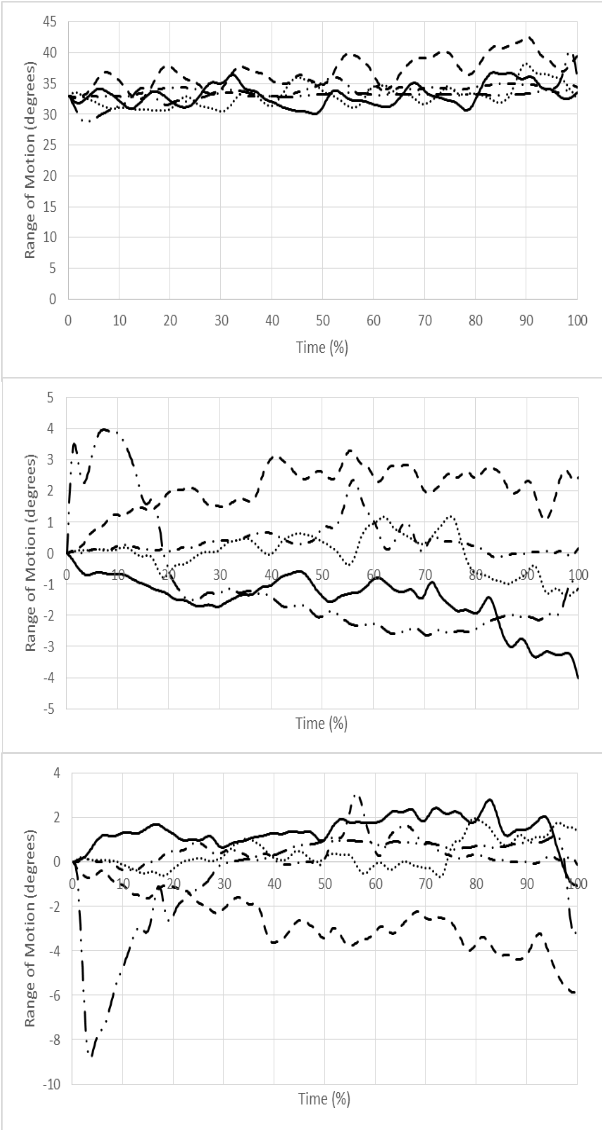

Figure 50 - Dynamic Upper Lumbar ROM for Machine Scrummaging

Table 51 - Maximum, Minimum and Peak-to-Peak Upper Lumbar ROM

	Flexion-Extension (°)	Lateral Bending (°)	Rotation (°)
Maximum	42	4	3
Minimum	29 (no –ve)	4	9
Peak-to-Peak	13	8	12

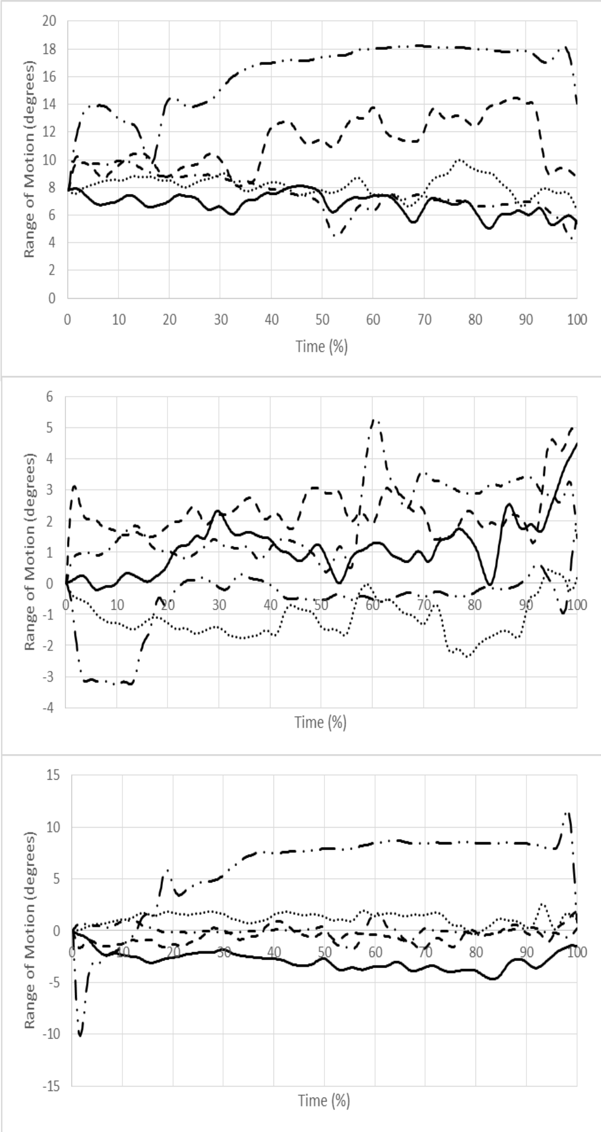

Figure 51 - Dynamic Lower Lumbar ROM for Machine Scrummaging

Table 52 - Maximum, Minimum and Peak-to-Peak Lower Lumbar ROM

	Flexion-Extension (°)	Lateral Bending (°)	Rotation (°)
Maximum	18	5	11
Minimum	4 (no –ve)	3	10
Peak-to-Peak	14	8	21

6.2.2.2. Live Scrummaging

The kinematic data presented in this section is for one particular participant performing both scrummaging sequences (CBS & CTPE) for each motion of all spinal segments. Each figure contains a number of different traces and these represent the spinal motion of a specific trial. For example, one figure may show cervical flexion-extension during live (CBS) scrummaging for all trials conducted for that participant. The data was time normalised so that the traces could be plotted on the same figure to demonstrate the variation. As stated in the previous chapter, this data represents data from a specific trial. That is, the solid black line represents the same trial throughout all dynamic ROM figures.

Qualitatively, it was difficult to identify any specific trends in the dynamic ROM traces, particularly for more mobile spinal segments, which would indicate that each individual scrummaging event is, to some extent, unique. It is therefore difficult to predict specific kinematic traces.

For all figures presented in this section, the format is as follows. The top two figures are flexion-extension for CBS and CTPE scrummaging respectively, the middle two figures are lateral bending for CBS and CTPE scrummaging respectively and the bottom two figures are rotation for CBS and CTPE scrummaging respectively. The tables presented show maximum, minimum and the peak-to-peak ROM across the trials presented.

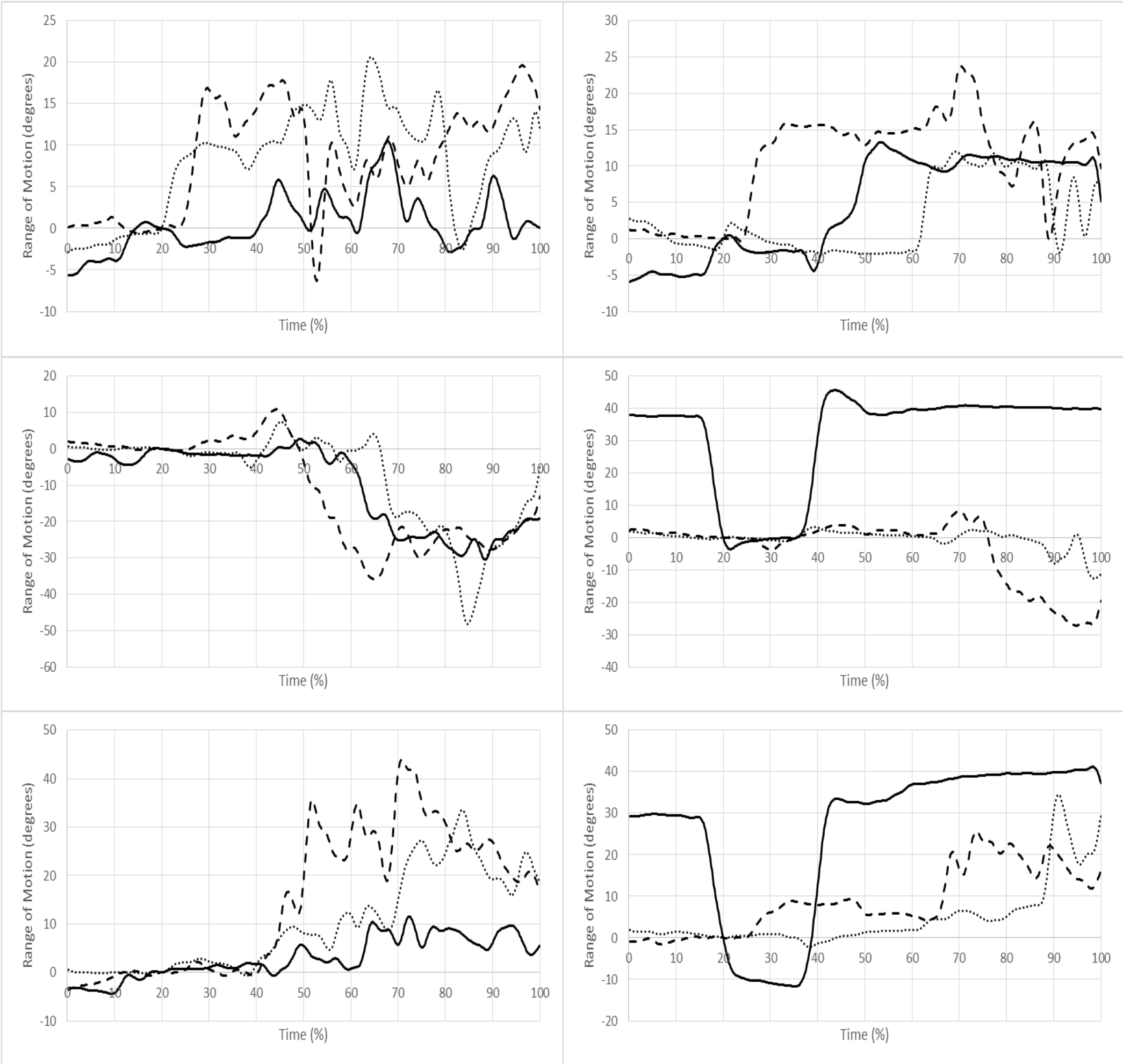

Figure 52 - Dynamic Cervical ROM for CBS and CTPE Live Scrummaging

Table 53 - Maximum, Minimum and Peak-to-Peak Cervical ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	CBS	**CTPE**	**CBS**	**CTPE**	**CBS**	**CTPE**
Maximum	21	23	11	45	43	41
Minimum	6	6	48	27	3	11
Peak-to-Peak	27	29	59	72	46	52

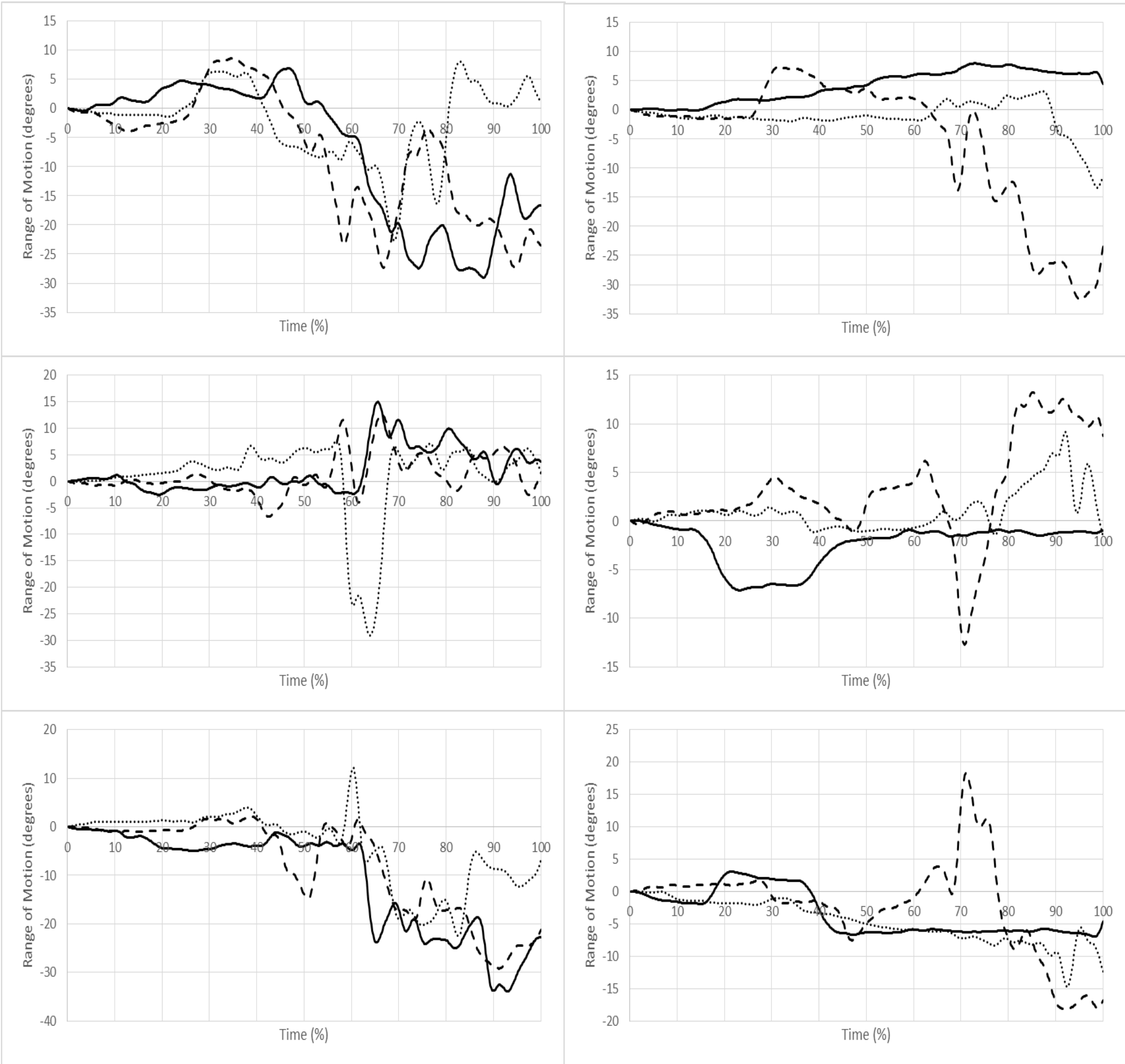

Figure 53 - Dynamic Upper Thoracic ROM for CBS and CTPE Live Scrummaging

Table 54 - Maximum, Minimum and Peak-to-Peak Upper Thoracic ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	CBS	**CTPE**	**CBS**	**CTPE**	**CBS**	**CTPE**
Maximum	9	8	15	13	121	17
Minimum	29	33	29	13	34	18
Peak-to-Peak	38	41	44	26	26	35

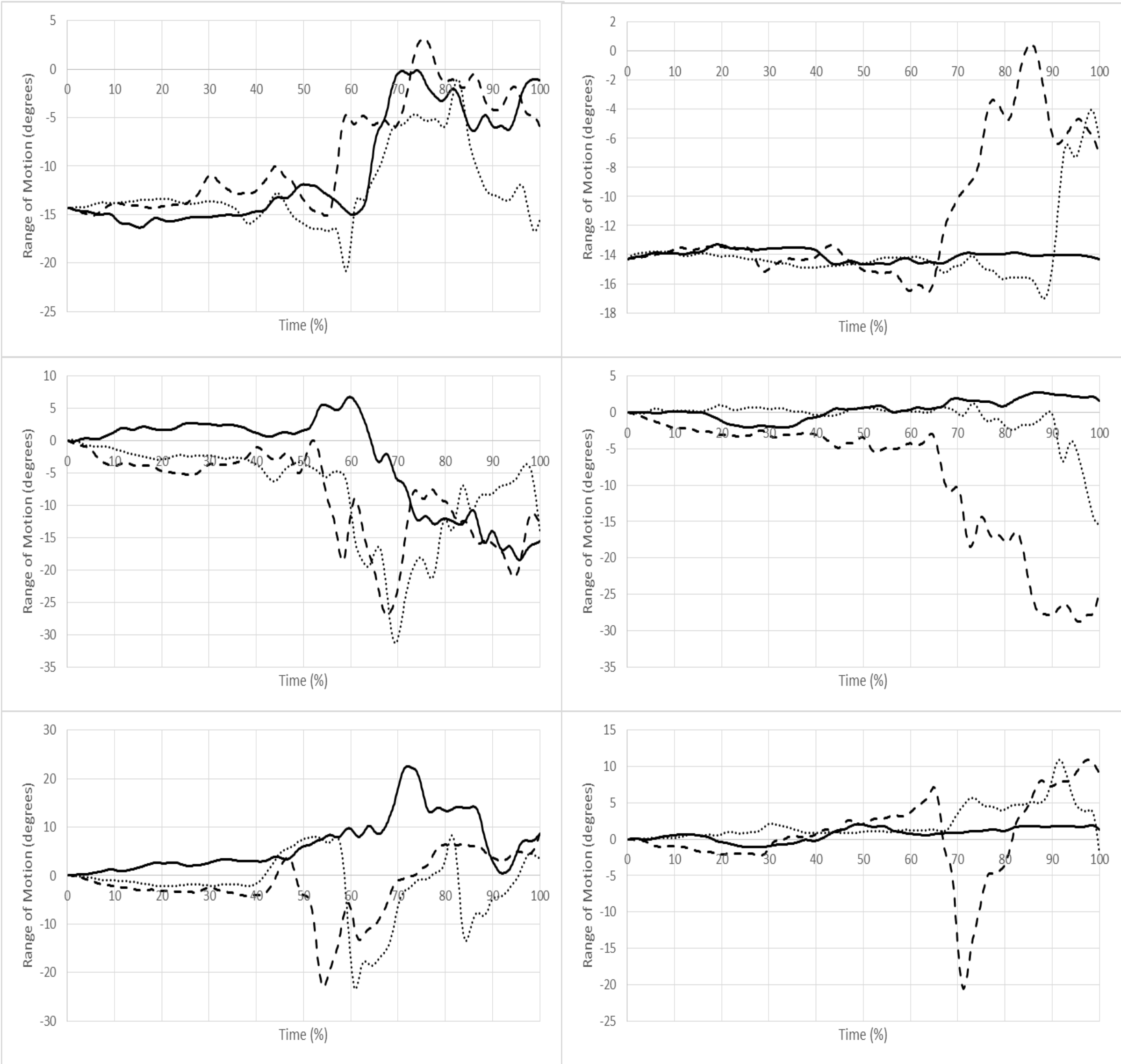

Figure 54 - Dynamic Lower Thoracic ROM for CBS and CTPE Live Scrummaging

Table 55 - Maximum, Minimum and Peak-to-Peak Lower Thoracic ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	CBS	**CTPE**	**CBS**	**CTPE**	**CBS**	**CTPE**
Maximum	3	<1	7	3	23	11
Minimum	21	17	31	28	23	21
Peak-to-Peak	24	17	38	31	46	32

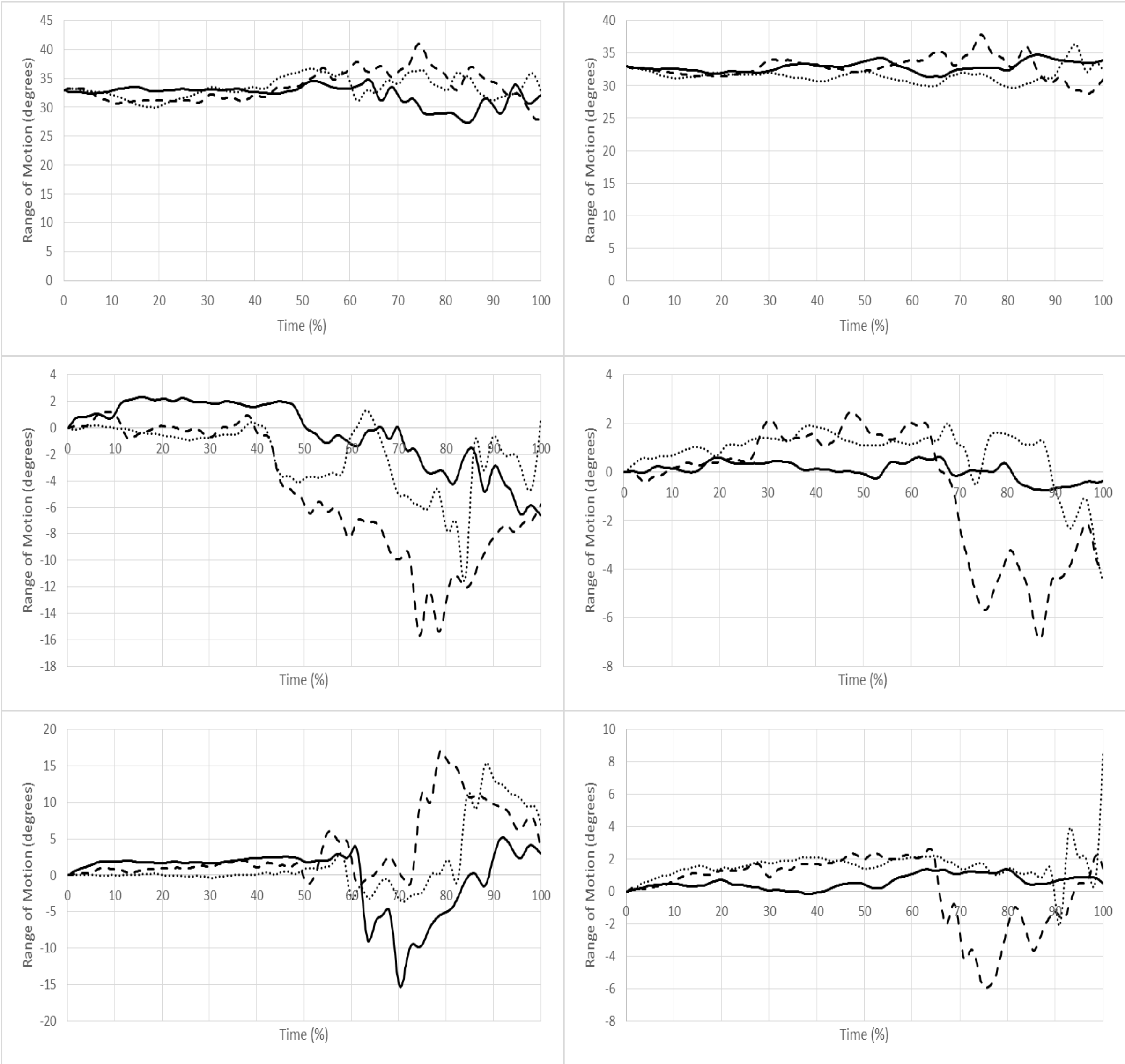

Figure 55 - Dynamic Upper Lumbar ROM for CBS and CTPE Live Scrummaging

Table 56 - Maximum, Minimum and Peak-to-Peak Upper Lumbar ROM

	Flexion-Extension (°)		**Lateral Bending (°)**		**Rotation (°)**	
	CBS	**CTPE**	**CBS**	**CTPE**	**CBS**	**CTPE**
Maximum	41	38	2	2	17	8
Minimum	27 (no –ve)	28 (no –ve)	16	7	15	6
Peak-to-Peak	14	10	18	9	32	14

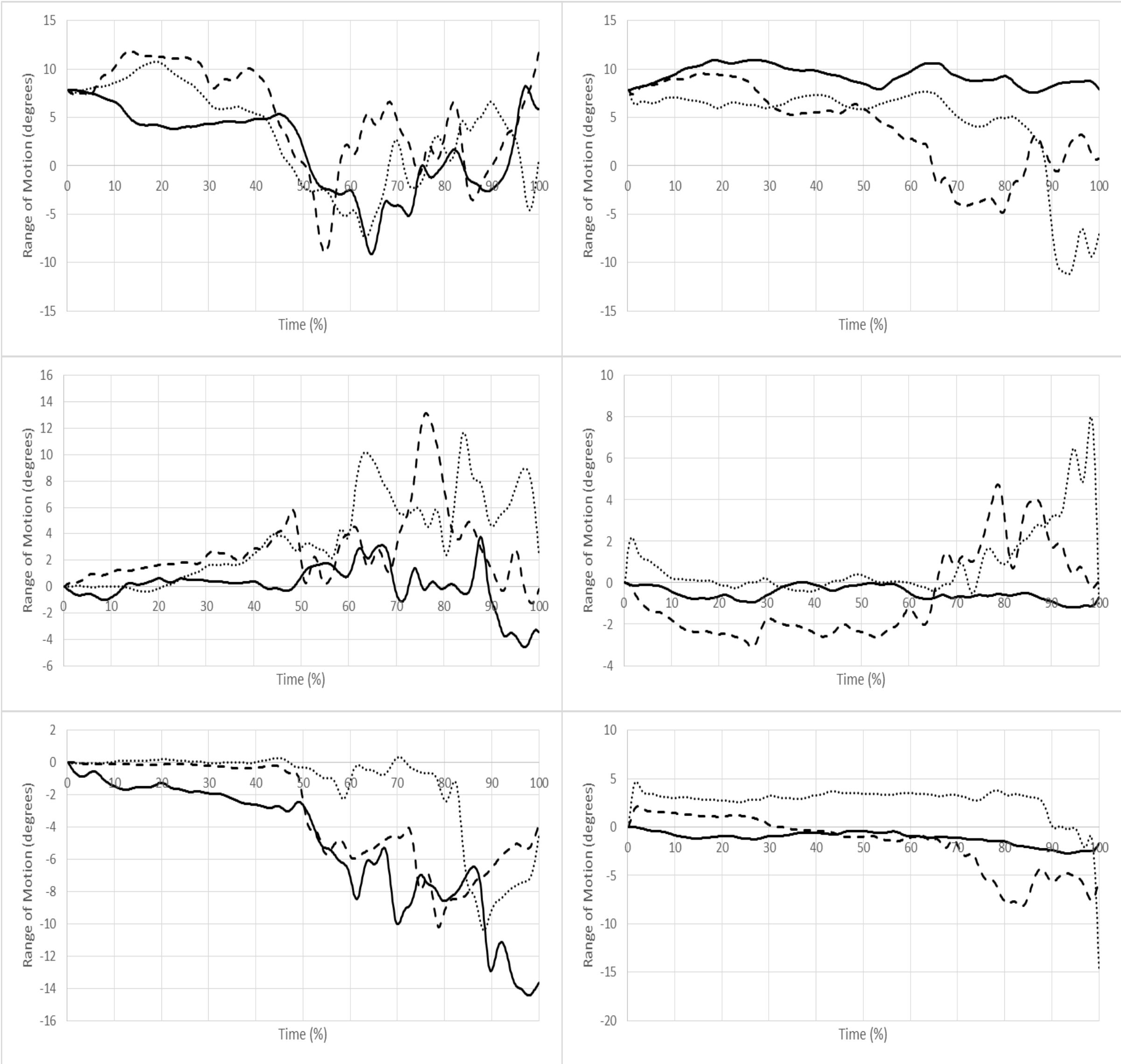

Figure 56 - Dynamic Lower Lumbar ROM for CBS and CTPE Live Scrummaging

Table 57 - Maximum, Minimum and Peak-to-Peak Lower Lumbar ROM

	Flexion-Extension (°)		Lateral Bending (°)		Rotation (°)	
	CBS	**CTPE**	**CBS**	**CTPE**	**CBS**	**CTPE**
Maximum	12	11	13	8	0	14
Minimum	9	11	5	3	14	5
Peak-to-Peak	21	22	18	11	14	9

6.2.2.4. Peak Kinematic Data

Table 58 shows mean peak cervical spine kinematics of all 14 participants that took part in all three scrummaging conditions. Mean data are presented with standard deviations in parentheses for both ROM and angular velocity. The repeated measures ANOVA revealed significant differences in cervical ROM for left lateral bending (p=0.011) and right rotation (p=0.021). After post-hoc pairwise comparisons, left lateral bending was significantly different when comparing machine (CBS) and live (CBS) scrummaging (p=0.02) and machine (CBS) and live (CTPE) scrummaging (p=0.04). Right rotation was significantly different between machine (CBS) and live (CTPE) scrummaging (p=0.013). For angular velocity, the repeated measures ANOVA revealed a significant difference in right lateral bending (p=0.017). After post-hoc pairwise comparisons, right lateral bending angular velocity was significantly different when comparing machine (CBS) and live (CBS) scrummaging (p=0.016) but not when comparing machine (CBS) and live (CTPE) scrummaging (p=0.343).

No significant differences (p>0.05) were observed for this segment for either live scrummaging technique (CBS vs CTPE). Differences in ROM between the two live (CBS vs CTPE) scrummaging were, at maximum 4°. Differences in angular velocity were, at maximum 3°/s but none of these differences proved to be statistically significant.

Table 58 – Mean Peak Cervical Spine Kinematic Data of the 14 Participants that took part in all Scrummaging Conditions. Mean peak data (degrees) are presented with standard deviation in parentheses.

	Range of Motion (°)			Angular Velocity (°/s)		
	Machine	**Live (CBS)**	**Live (CTPE)**	**Machine**	**Live (CBS)**	**Live (CTPE)**
Flexion	11.06 (8.19)	18.11 (9.26)	16.56 (10.60)	7.21 (2.19)	9.02 (4.97)	6.54 (2.45)
Extension	18.17 (5.46)	17.79 (9.91)	16.90 (11.19)	-6.26 (2.13)	-11.13 (6.90)	-8.18 (3.65)
Right Lateral Bending	16.14 (8.66)	18.54 (9.82)	20.82 (11/28)	5.39 (2.56)	10.18 (6.27)	8.98 (7.08)
Left Lateral Bending	9.93 (5.32)	19.49 (12.25)*	19.15 (13.33)*	-6.05 (1.76)	-7.17 (2.46)	-8.94 (3.79)
Right Rotation	14.27 (9.74)	20.88 (9.11)	24.93 (10.06)*	8.53 (3.79)	6.93 (5.35)	9.89 (7.12)
Left Rotation	17.36 (9.08)	12.75 (7.81)	14.82 (10.62)	-6.93 (5.34)	-9.55 (4.65)	-7.92 (4.57)

* - denotes a significant difference (p<0.05) between the variable marked and machine (CBS) scrummaging

Table 59 shows mean peak upper thoracic spine kinematics of all 14 participants that took part in all three different scrummaging conditions. Mean data are presented with standard deviations in parentheses for both ROM and angular velocity. The repeated measures ANOVA revealed significant differences in ROM for right lateral bending (p=0.010). The post-hoc pairwise comparisons revealed that these differences were present between machine (CBS) and live (CBS) scrummaging (p=0.007) as well as between machine (CBS) and live (CTPE) scrummaging (p=0.010). For angular velocity, no significant differences were revealed.

No significant differences were observed for the live scrummaging sequences for this segment (p>0.05). Differences in mean peak ROM between the two live (CBS vs CTPE) scrummaging sequences for this segment were, at maximum, 5.5° and differences in angular velocity were, at maximum, 2.5°/s but none of the kinematic differences were statistically significant. This indicates the similarity in mean peak kinematic variables for this spinal segment for the two live scrummaging sequences.

Table 59 – Mean Peak Upper Thoracic Spine Kinematic Data of the 14 Participants who took part in all Scrummaging Conditions. Mean peak data (degrees) are presented with standard deviation in parentheses.

	Range of Motion (°)			Angular Velocity (°/s)		
	Machine	**Live (CBS)**	**Live (CTPE)**	**Machine**	**Live (CBS)**	**Live (CTPE)**
Flexion	9.00 (5.34)	8.92 (7.30)	10.61 (5.37)	4.98 (1.64)	10.09 (6.82)	10.25 (7.86)
Extension	13.58 (5.30)	22.10 (10.04)	25.15 (13.51)	-5.71 (1.90)	-5.87 (2.97)	-6.69 (2.41)
Right Lateral Bending	22.10 (10.04)	10.56 (6.18)*	11.38 (6.84)*	3.98 (1.34)	7.24 (4.72)	7.20 (5.11)
Left Lateral Bending	25.15 (13.51)	15.33 (11.77)	15.83 (9.88)	-4.85 (2.73)	-7.18 (3.12)	-6.33 (2.30)
Right Rotation	10.59 (5.67)	14.74 (12.50)	10.70 (7.33)	8.18 (10.55)	9.93 (9.64)	7.42 (2.74)
Left Rotation	14.74 (12.50)	16.02 (9.58)	21.68 (10.10)	-6.59 (4.40)	-10.71 (10.59)	-9.26 (6.79)

* - denotes a significant difference ($p<0.05$) between the variable marked and machine (CBS) scrummaging

Table 60 shows mean peak lower thoracic ROM during all three scrummaging conditions investigated. Mean data are presented for both ROM and angular velocity with standard deviations in parentheses. For ROM, the repeated measures ANOVA revealed significant differences for left lateral bending ($p<0.001$) with post-hoc pairwise comparisons revealing differences between machine (CBS) and live (CBS) scrummaging ($p=0.019$) and between machine (CBS) and live (CTPE) scrummaging ($p<0.001$). For angular velocity, the repeated measures ANOVA did not reveal any significant differences ($p>0.05$).

No significant differences ($p>0.05$) were observed for any kinematic data between the two live (CBS vs CTPE) scrummaging techniques. Differences in mean peak lower thoracic ROM were, at maximum, 4°. Differences in angular velocity were, at maximum, 1.4°/s. The non-statistical difference between the two live scrummaging sequences indicate similar kinematics for this spinal segment.

Table 60 – Mean Peak Lower Thoracic Spine Kinematic Data of the 14 Participants who took part in all Scrummaging Conditions. Mean peak data (degrees) are presented with standard deviation in parentheses.

	Range of Motion (°)			Angular Velocity (°/s)		
	Machine	**Live (CBS)**	**Live (CTPE)**	**Machine**	**Live (CBS)**	**Live (CTPE)**
Flexion	3.97 (4.85)	4.67 (6.65)	5.12 (5.40)	4.18 (1.90)	4.38 (2.29)	5.74 (5.15)
Extension	16.10 (6.26)	15.98 (8.98)	17.42 (10.34)	-4.38 (2.64)	-3.84 (2.64)	-4.22 (2.49)
Right Lateral Bending	8.49 (6.37)	6.74 (3.98)	7.62 (4.70)	4.40 (2.77)	6.02 (3.96)	6.11 (3.14)
Left Lateral Bending	7.95 (3.79)	15.29 (7.42)*	16.57 (5.49)*	-4.61 (2.49)	-4.85 (1.98)	-5.13 (1.71)
Right Rotation	6.83 (3.71)	14.03 (11.64)	9.93 (7.24)	5.16 (4.18)	5.55 (3.75)	5.40 (3.60)
Left Rotation	9.36 (6.47)	11.98 (5.77)	13.18 (6.76)	-5.77 (3.89)	-8.26 (4.89)	-7.81 (5.55)

* - denotes a significant difference ($p<0.05$) between the variable marked and machine (CBS) scrummaging

Table 61 shows mean peak upper lumbar ROM and angular velocity during all three conditions with standard deviations in parentheses. The repeated measures ANOVA did not reveal any significant differences ($p>0.05$) in upper lumbar ROM. For angular velocity, no significant differences ($p>0.05$) were present.

No significant differences ($p>0.05$) were observed for any comparison between the two live (CBS vs CTPE) scrummaging techniques. Differences in upper lumbar ROM were, at maximum, 3.2° and differences in angular velocity were, at maximum, 3.5°/s. No statistically significant differences were observed between the two live scrummaging engagement sequences thus indicating the similarity in upper lumbar kinematics.

Table 61 – Mean Peak Upper Lumbar Spine Kinematic Data of the 14 Participants who took part in all Scrummaging Conditions. Mean peak data (degrees) are presented with standard deviation in parentheses.

	Range of Motion (°)			Angular Velocity (°/s)		
	Machine	**Live (CBS)**	**Live (CTPE)**	**Machine**	**Live (CBS)**	**Live (CTPE)**
Flexion	42.36 (10.33)	42.32 (10.99)	43.11 (12.27)	4.03 (1.86)	5.02 (2.53)	4.98 (2.48)
Extension	0.75 (1.58)	0.36 (0.83)	0.81 (3.05)	-4.14 (2.60)	-4.19 (2.60)	-4.65 (3.36)
Right Lateral Bending	6.87 (6.10)	7.25 (4.76)	7.90 (5.98)	4.29 (2.60)	5.07 (2.44)	4.61 (2.47)
Left Lateral Bending	7.55 (4.33)	10.64 (6.34)	10.90 (6.25)	-3.85 (2.30)	-5.34 (4.31)	-4.80 (3.02)
Right Rotation	9.73 (9.91)	12.89 (7.93)	13.48 (5.88)	4.87 (4.68)	5.62 (5.10)	6.04 (4.56)
Left Rotation	6.40 (4.64)	7.98 (6.83)	7.67 (7.41)	-4.09 (3.71)	-6.66 (5.09)	-9.12 (12.21)

* - denotes a significant difference ($p<0.05$) between the variable marked and machine (CBS) scrummaging

Table 62 shows mean peak ROM and angular velocity data for the lower lumbar segment for all three conditions investigated with standard deviations in parentheses. The repeated measures ANOVA revealed no significant differences ($p>0.05$) for any mean peak ROM variables of the lower lumbar segment. Similarly, for angular velocity, the repeated measures ANOVA did not reveal any significant differences ($p>0.05$).

No significant differences ($p>0.05$) were observed between the two live scrummaging sequences for any kinematic variables of this segment. Differences in lower lumbar ROM were, at maximum, 4° but none of these were statistically significant. For angular velocity, differences were, at maximum, 2°/s. As with the other spinal segments, since no statistically significant differences were observed, the data suggests similar peak lower lumbar kinematics between the two live scrummaging sequences.

Table 62 – Mean Peak Lower Lumbar Spine Kinematic Data of the 14 Participants who took part in all Scrummaging Conditions. Mean peak data (degrees) are presented with standard deviation in parentheses.

	Range of Motion (°)			Angular Velocity (°/s)		
	Machine	**Live (CBS)**	**Live (CTPE)**	**Machine**	**Live (CBS)**	**Live (CTPE)**
Flexion	11.27 (12.01)	16.11 (17.09)	14.17 (12.54)	4.25 (3.06)	4.92 (3.69)	5.13 (3.14)
Extension	9.95 (5.57)	11.56 (9.08)	9.77 (7.38)	5.01 (3.75)	8.06 (6.16)	6.50 (3.76)
Right Lateral Bending	7.24 (6.26)	9.37 (6.68)	7.48 (5.06)	3.90 (2.12)	5.62 (4.61)	3.75 (1.99)
Left Lateral Bending	7.93 (6.74)	10.09 (9.03)	8.32 (6.44)	4.19 (3.52)	4.47 (2.95)	4.16 (1.45)
Right Rotation	10.11 (8.55)	9.97 (10.94)	12.65 (10.08)	5.67 (5.86)	7.05 (5.30)	6.77 (5.57)
Left Rotation	10.01 (8.66)	15.88 (11.96)	11.89 (10.27)	5.25 (4.84)	8.08 (6.86)	6.06 (3.43)

* - denotes a significant difference ($p<0.05$) between the variable marked and machine (CBS) scrummaging

6.2.2.5. Kinematic Data Variation

Table 63 - Table 66 show the coefficient of variation for all segments and motions of four different participants. Data for all other participants can be found in the appendices. For participant 1 (Table 63), the coefficient of variation for peak-to-peak ROM ranges from 1.6-51.8% which both occur during live (CBS) scrummaging. The minimum variation was for rotation of the upper lumbar segment and the maximum variation was for flexion-extension of the lower thoracic segment. For participant 8 (Table 64) the coefficient of variation ranges from 3.1-67.2%. The minimum variation was for upper thoracic rotation during live (CBS) scrummaging and the maximum variation was for lower lumbar lateral bending during live (CTPE) scrummaging and lower lumbar rotation during machine scrummaging. For participant 9 (Table 65), the coefficient of variation ranges from 1.8-63.5%. Minimum variation occurred for lateral bending of the upper thoracic segment during live (CTPE) scrummaging and maximum variation occurred for lower lumbar lateral bending during live (CBS) scrummaging. For participant 10 (Table 66), the coefficient of variation ranged from

4.3-99.5%. Minimum variation was for upper thoracic rotation during live (CBS) scrummaging and maximum variation was for lower lumbar rotation during machine scrummaging.

From this data, it can be seen that regardless of the condition, peak-to-peak ROM can have a large variation. Furthermore, an ANOVA was used to determine whether any significant differences were present between any of the scrummaging conditions in terms of the variation in peak-to-peak ROM for all participants. It was determined that there were no significant differences ($p>0.05$) in the variation in the data between any of the conditions.

Table 63 - Coefficient of Variation for all Segments and all Conditions for Participant 1

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	33.4	12.0	5.1	26.6	28.1	24.0	18.7	29.0	42.3	27.7	45.6	41.6	32.9	29.2	38.9
Live (CBS)	51.1	12.4	26.2	41.4	24.1	21.3	51.8	14.1	15.4	10.5	18.7	1.6	8.9	26.6	15.2
Live (CTPE)	19.0	32.6	18.3	23.8	21.1	19.0	7.3	9.8	9.3	7.3	19.0	33.2	21.7	39.6	18.5

Table 64 – Coefficient of Variation for all Segments and all Conditions for Participant 8

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	28.3	26.8	59.9	34.7	33.3	24.4	39.3	28.7	23.1	40.6	43.7	26.7	36.5	66.1	67.2
Live (CBS)	11.5	14.2	25.3	40.0	17.6	3.1	2.9	39.1	6.4	13.0	18.4	41.5	6.2	28.2	3.0
Live (CTPE)	40.7	29.0	26.3	5.5	26.9	6.1	28.7	11.8	40.8	16.0	28.2	57.3	9.3	67.2	43.4

Table 65 – Coefficient of Variation for all Segments and all Conditions for Participant 9

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	38.4	37.8	27.8	53.4	8.7	32.3	35.7	21.5	42.9	30.7	36.5	29.6	47.8	53.2	56.2
Live (CBS)	23.1	28.7	21.9	16.9	19.5	16.1	19.9	11.1	21.1	14.4	63.3	52.2	27.4	63.5	49.1
Live (CTPE)	3.7	4.7	30.0	19.5	1.8	22.3	37.8	30.2	6.4	22.7	25.3	56.5	58.3	59.7	54.9

Table 66 – Coefficient of Variation for all Segments and all Conditions for Participant 10

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	29.8	53.5	22.9	26.5	50.0	69.0	28.3	46.7	44.2	36.4	39.9	51.2	43.6	20.1	99.5
Live (CBS)	19.1	20.5	39.8	7.2	35.4	4.3	8.0	9.4	15.0	31.8	25.2	4.4	7.7	31.0	37.7
Live (CTPE)	21.3	40.7	27.1	62.7	55.8	56.1	63.4	58.3	74.0	36.0	57.5	56.4	53.5	57.0	63.3

6.2.2. EMG and Force Data

EMG and force data was collected during machine scrummaging and used to create an individualised correlation curve to then use as a means of predicting muscle force production during live scrummaging. This method was validated in Chapter 2 in a laboratory environment. The limitations of this method in a dynamic environment are acknowledged by the author and these are discussed in a later section. Figure 57 shows vertical force production during machine scrummaging at the requested exertions of minimum, 25% MVC, 50% MVC, 75% MVC and 100% MVC. Each of the peaks were easily identifiable to input into the EMG and force correlation curve.

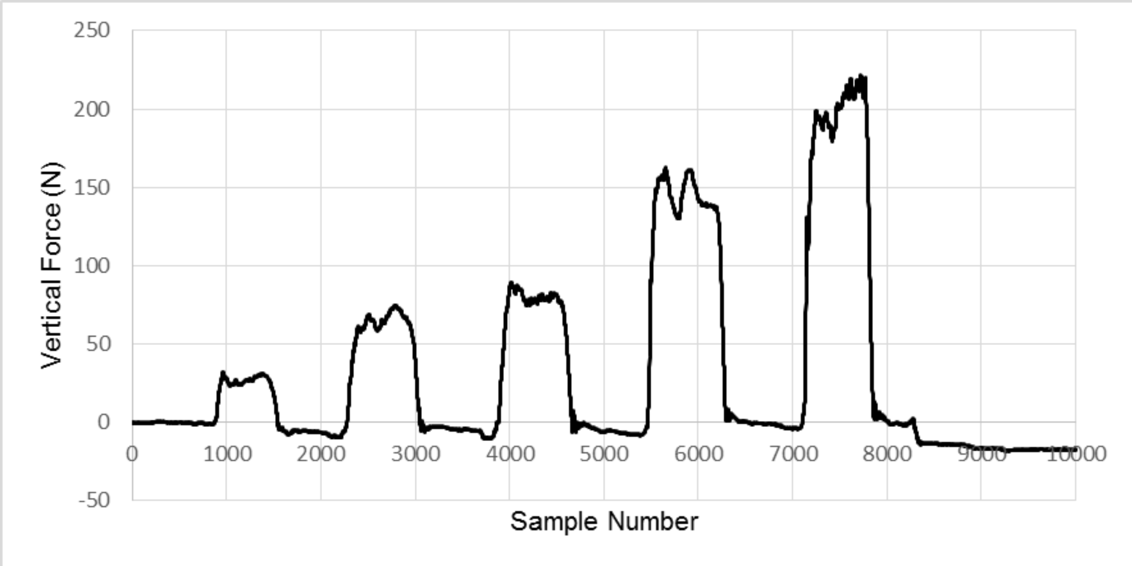

Figure 57 – Example of a machine scrummaging trial of gradually increasing force from minimum to maximum

Figure 58 shows an example corresponding to Figure 57 of the left and right CES muscle activity during the gradually increasing exertion trials. The peaks corresponding to the same time intervals as the force peaks were identified and then input into the force-EMG correlation curve. This type of trial was repeated 3 times in order to obtain 15 data points for each individualised correlation curve.

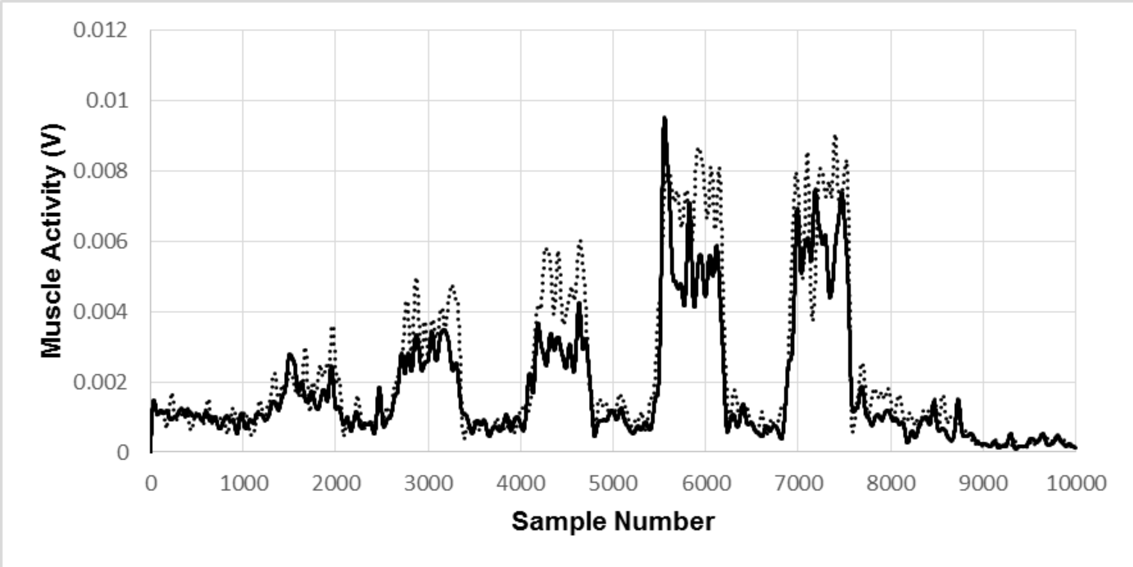

Figure 58 – Example of increasing activity during machine scrummaging corresponding to Figure 57. Solid line - Right CES Activity; Dotted line - Left CES Activity

Figure 59 shows an example of an individualised force-EMG correlation curve. During preliminary data analysis, polynomial trends were also fitted to the data but did not result in as strong correlation coefficient values. Second degree polynomials resulted in R^2 values ranging from 0.65-0.90 for all participants and third degree polynomials resulted in R^2 values of 0.6-0.8. A linear fit resulted in R^2 values of 0.70-0.95 and thus this trend was fitted to the data as it resulted in the strongest correlation.

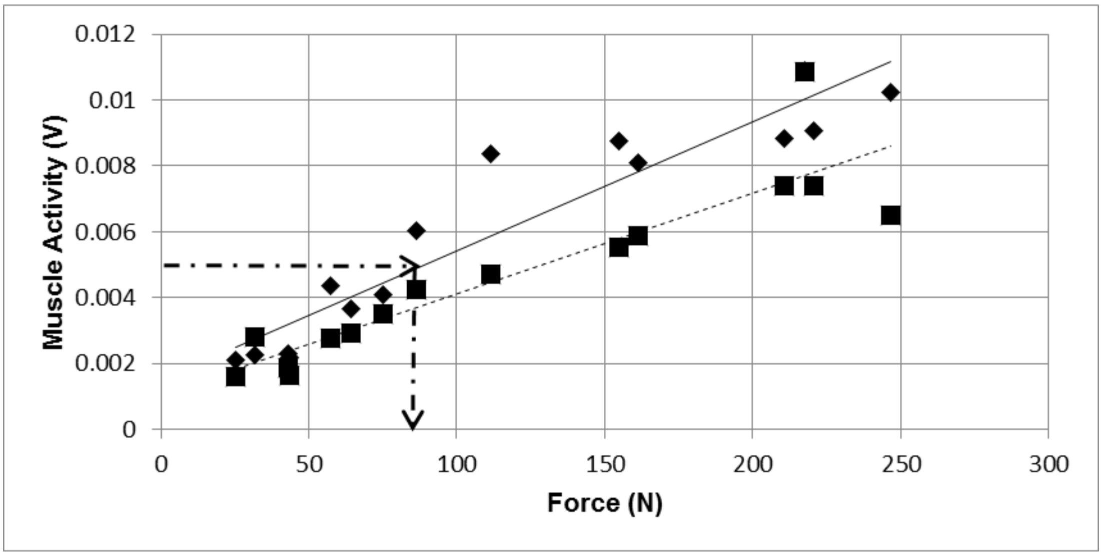

Figure 59 – Example of a Force-EMG correlation curve from three gradually increasing exertion trials. Diamonds – Left Side; Square – Right Side. Solid line – Left side trend line, R^2=0.945; Dashed line – Right side trend line, R^2=0.911

Figure 60 shows example EMG traces of the left and right CES musculature during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for one trial of each condition for one participant. During machine scrummaging (Figure 60(a)), there is a maximum for both left and right CES muscle activity (0.010V and 0.007V respectively) as the participant engages with the scrum machine and drives upwards. Both left and right CES muscles then reduced in activity during the sustained scrummaging phase with maximums of 0.008V and 0.006V respectively before reducing back to a much lesser amount (0.002V) as the pack disengages with the machine. During live scrummaging (CBS; Figure 60(b)), there is a maximum in CES muscle activity for the right side on engagement (0.008V) but there is no maximum activity for the left side at the same time point. Left CES muscle activity gradually increases throughout the trial to a maximum of 0.0065V and this then reduces as the scrum disengages. The right side activity fluctuates a large amount but there is a concurrent peak with left CES muscle activity towards the end of the trial at 0.0055V. There is then much less activity for both muscles as the scrum disengages (0.001V). Finally, during live scrummaging (CTPE; Figure 60(c)), there is an increase in muscle activity as the scrum engages but there is not a sharp increase as can be seen during the previous two conditions. There is a gradual increase towards the middle of the trial for the right side with a maximum of (0.006V) and this then decreases towards the end of the trial (0.004V) before the muscle activity decreases as the scrum disengages. For the left side, there is a gradual increase in muscle activity throughout the trial to a maximum of 0.003V and this then decreases as the scrum disengages. For both the live scrummaging conditions, it appears that this player favours bearing the forced flexion with the right sided musculature which is likely to cause asymmetric spinal loading. For this participant, the maximum EMG values equated to vertical forces of (a) 381N, (b) 233N and (c) 114N after using the adjustment coefficient developed in Chapter 3.

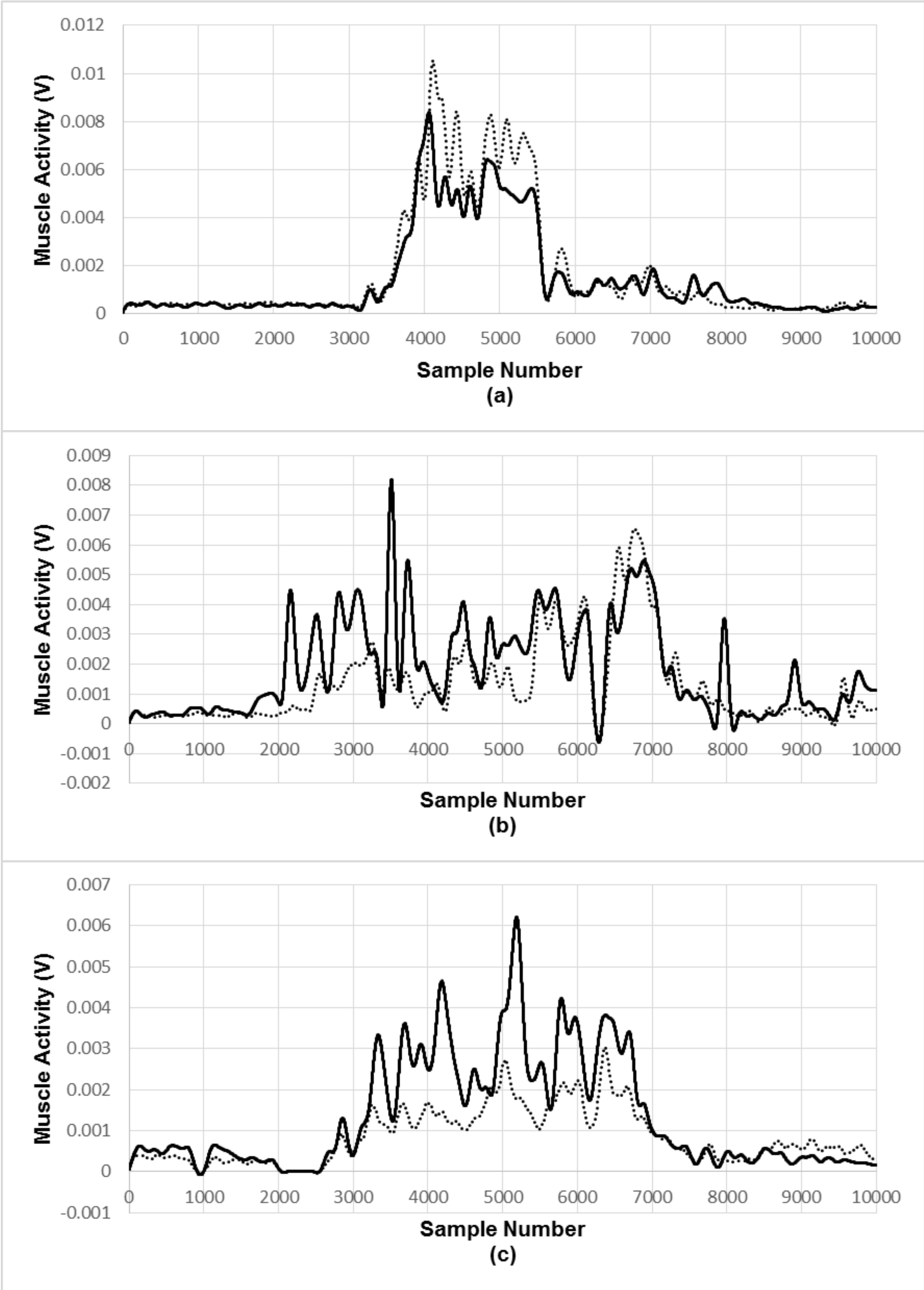

Figure 60 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging. Solid line - Right CES; Dotted line - Left CES

Table 67 shows vertical force production data for all participants who took part in the force-EMG trials of all three scrummaging conditions. During machine scrummaging, the vertical force production measured via the inverted force platform ranged from 180-309N for the 9 participants that took part in this part of the trial. During live (CBS) scrummaging, the determined vertical force production ranged from 145-329N and during live (CTPE) scrummaging, the determined vertical force production ranged from 163-369N. Standard deviation for the group was calculated using the mean maximum vertical force production for each participant. Standard deviations for each participant were calculated from the maximum vertical force determined from each of the trials.

Table 67 – Mean Maximum Vertical Force Production for all Participants who took part in the Force-EMG trials. Mean Data are Presented

Participant No.	Machine	Live (CBS)	Live (CTPE)
1	232.7 (11.4)	145.3 (38.4)	163.1 (40.6)
2	180.5 (5.8)	152.0 (78.1)	174.5 (24.9)
3	175.2 (10.2)	243.3 (22.8)	173.0 (17.9)
4	225.3 (7.8)	271.1 (49.2)	208.9 (23.7)
5	224.7 (40.1)	199.2 (25.8)	221.9 (8.3)
6	223.8 (7.1)	N/A	N/A
7	252.4 (41.4)	329.0 (18.8)	369.4 (19.4)
8	309.2 (28.6)	225.6 (22.3)	274.3 (39.5)
9	215.2 (15.6)	263.3 (33.6)	N/A
Mean (SD)	226.9 (39.4)	228.6 (58.1)	226.4 (68.3)

N/A – data for EMG could not be obtained or was extremely noisy and thus determined vertical force values could not be obtained.

The repeated measures ANOVA revealed that no significant differences were present for any comparison of maximum EMG and (determined) force data. This would indicate that these players do not significantly alter their scrummaging technique according to the differing scrummaging conditions that were investigated or that the two live scrummaging sequences do not have a significant impact on the players' cervical erector spinae muscle activity magnitude.

6.2.3. Scrum Collapse Kinematics

This section seeks to explore some of the kinematic data that occurred during the scrum collapse. Only one scrum collapse occurred for each engagement sequence (CBS/CTPE) throughout the trials during this study.

Figure 61 shows dynamic cervical (top) and upper lumbar (bottom) ROM during scrum collapse of the CBS sequence. The collapse occurred at the end of the trial just prior to sample number 500. The cervical spine is in a position of 30° flexion, 21° right lateral bending and 21° right rotation simultaneously. None of these values exceeded the maximum ROM for this individual although it does equate to 80% flexion, 60% lateral bending and 40% of rotation. If the participant does not exceed their maximum ROM, it is unlikely an injury would occur. Combined with the resulting force provided by the second and back rows of the scrum, however, it may be suggested that this scenario provides an increased risk of injury for the player in question. In vitro studies have shown the strength of the cervical spine to be less resistant to compressive force when in flexion (Przybyla et al. 2007) which would suggest an increased likelihood of spinal failure when in a flexed position. The upper lumbar spine ROM data is given to determine whether a possible change in lumbar spine kinematics pre-empts the collapse of the scrum. During this particular event, just prior to collapse, the upper lumbar spine increases its extent of flexion with a maximum of 34° and a large amount of right rotation (52°) is also present. It is difficult to determine whether this motion of the upper lumbar spine is a precursor to scrum collapse but it is possible that there may be a link.

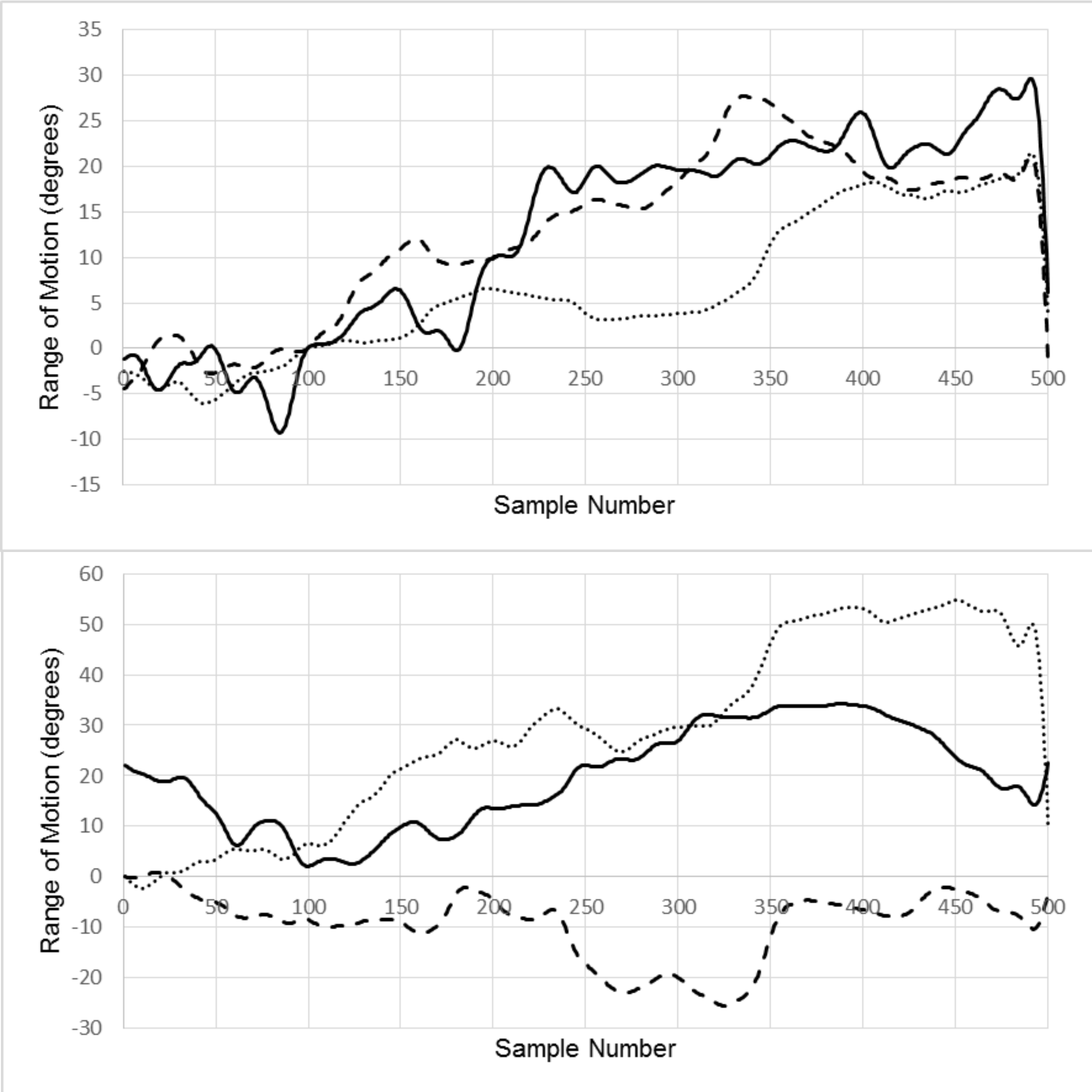

Figure 61 – Dynamic Cervical (top) and Upper Lumbar (bottom) ROM during live scrum collapse (CBS). Solid line – Flexion-Extension; Dashed Line – Lateral Bending; Dotted Line – Rotation

Figure 62 shows dynamic cervical (top) and upper lumbar (bottom) ROM during scrum collapse of the CTPE engagement sequence. Once again, the collapse occurred at the end of the trial and it can be observed that large amounts of cervical spine ROM were utilised by this participant. The cervical spine is forced into a position of 64° flexion, 65° left lateral bending and 31° left rotation. Again, the participant did not exceed their maximum ROM in any plane although this did equate to a large amount of available ROM in each plane. This equated to 95% in the sagittal plane (flexion), 95% of lateral bending and 60% of available left rotation. The participant again did not exceed the maximum available ROM and so it is unlikely they are to be injured. Combined with the force from the other players of the scrum as the player's head is driven into the turf does, however, put this player at additional risk.

As previously stated, when in non-neutral positions, the resistance of the cervical spine to compressive force is reduced and this has been shown in vitro (Przybyla et al. 2007). The upper lumbar ROM has also been presented in this case to determine whether scrum collapse is pre-empted by large amounts of motion in this segment. During this particular collapse, however, there is an increase to 30° flexion which decreases to 20° and then increases to 27° just prior to collapse. In this case of scrum collapse, there is not a great magnitude of motion used in any plane for this segment prior to collapse which would suggest that there is not a relationship between upper lumbar kinematics and collapse.

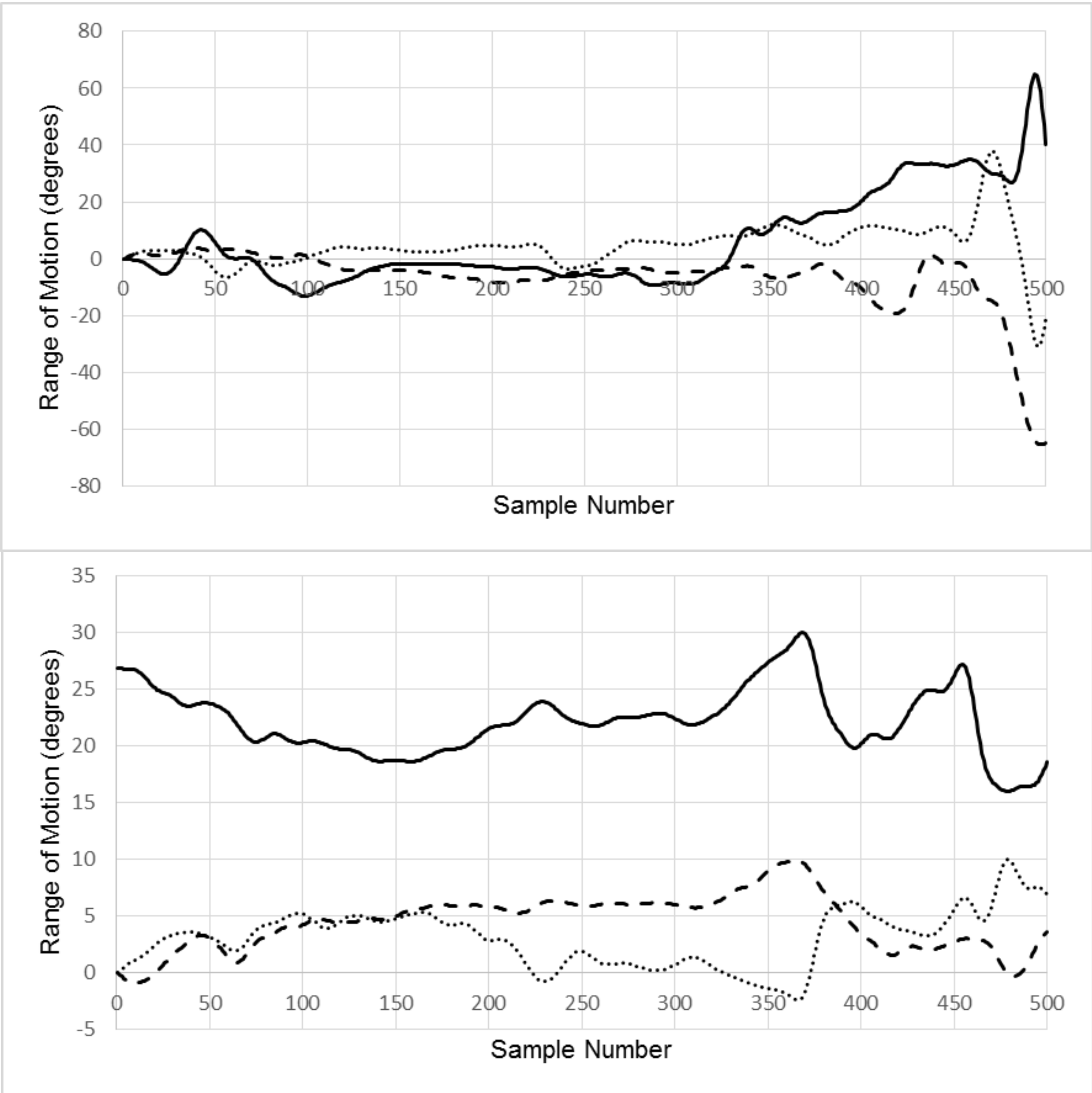

Figure 62 – Dynamic Cervical (top) and Upper Lumbar (bottom) ROM during live scrum collapse (CTPE). Solid line – Flexion-Extension; Dashed Line – Lateral Bending; Dotted Line – Rotation

6.3. Summary of Results

The data presented in this study indicates that machine scrummaging represented a more constrained biomechanical environment when considering peak ROM, given that the machine scrummaging ROM data rarely exceeded the equivalent live data. Indeed, there were some statistically significant differences ($p<0.05$) when considering peak ROM when comparing machine and live environments. There was, however, no significant difference ($p>0.05$) between machine and live scrummaging when considering peak muscle activity and peak determined force.

When comparing the two live scrummaging conditions (CBS vs CTPE), no significant differences ($p>0.05$) were found for peak kinematic data, peak muscle activity and peak determined force. This would suggest that the change in scrum engagement laws has little effect on the hooker when considering these variables.

Once again, the variation in peak-to-peak ROM of the five spinal segments in each plane was not consistent across participants and the different experimental conditions. This would suggest that each individual reacts and copes with the unique biomechanical loading scenario of the scrum in an individual way. Thus, a change in the laws governing the scrum may not provide a more favourable biomechanical scenario when considering these variables as previous research would suggest (Cazzola, Preatoni, et al. 2015; Cazzola et al. 2014b).

Finally, this study provided an insight into scrum collapse kinematics for the hooker; one for each engagement sequence. A relatively large amount of coupled motion was observed for each scenario with each participant being exposed to almost all of their physiological ROM. Their physiological ROM was not exceeded, but combined with the force produced from the second and back row players as the scrum collapsed, it is likely that the hooker is exposed to a scenario where the risk of injury is high.

6.4. Conclusions

This study has been unable to provide stand-alone evidence that the new rugby union scrum engagement laws positively enhances the hooker’s spinal kinematics. The consistency of data across live scrummaging conditions presented here indicates that the CBS sequence is unlikely to represent an improved biomechanical scenario when considering spinal kinematics and muscle activity of the cervical erector spinae musculature.

7. General Discussion

This thesis set out with the aim of developing a method to assess spinal biomechanics during rugby scrummaging and address the overall research question of how the evolution of rugby affects the risk of spinal injury for the hooker. This was achieved by evaluating a variety of kinematic measurement techniques, developing a method to indirectly measure neck muscle force production and finally, analysing the effect of changing the playing surface and engagement sequence on hooker spinal biomechanics.

7.1. Inertial Sensor Validation

Absolute errors between orientation measurements of the inertial sensors and the digital encoders of the rotary table ranged between 3.20° (2.20%) for the pitch axis and 14.31° (3.97%) for the heading axis (Table 9). Lin's coefficient of concordance was used in order to determine how accurate the 1:1 relationship between inertial sensor and digital encoder output was. Similar to correlation, values greater than 0.8 are considered high. For each sensor and each axis, the minimum value of concordance was 0.987 and the maximum was 0.999 indicating an almost perfect 1:1 relationship of orientation between the two devices.

Compared to other motion analysis systems, the absolute error of these inertial sensors was slightly higher. Electromagnetic tracking systems have demonstrated average errors of 1.5° (Hassan et al. 2007) and opto-electronic systems have been shown to have errors of ±2° (Pearcy et al. 1987). These inertial sensors demonstrated absolute errors, at maximum, of 3.2° which is 50% greater than other motion analysis systems. This error must be taken into account when interpreting the data but this error was considered acceptable as using this instrumentation provided a cost-effective and feasible solution to implement during rugby scrummaging. Furthermore, when considering the large amount of variation in the data, this level of error is unlikely to have a large effect on the ROM values presented. Sources of error of inertial sensor orientation are gyroscopic drift and the interference of any ferrous metals which will affect the output of the magnetometer. These sensors, however, were integrated with a fusion algorithm which incorporated Kalman filtering to provide data that was free of drift. This type of filter has been shown to be the most appropriate method for the correction of gyroscopic drift (de Vries et al. 2009).

Despite the errors associated with this set of inertial sensors, they provided a solution for the measurement of spinal kinematics during rugby scrummaging that other systems did not. Therefore, with the high values of concordance shown in the laboratory validation, they were deemed an appropriate method for the application they were required for.

7.2. Force-EMG Study

This laboratory-based study demonstrates the use of EMG to predict muscle force in the cervical region. The results demonstrate no significant difference between the actual force and that determined by EMG. The high correlation coefficient reported in this study provides evidence of a linear relationship between CES force and EMG activation (Figure 4.6), which is consistent with previous studies considering other muscles groups (Keshner and Campbell 1989; Queisser et al. 1994). It is, however, in contradiction of other studies where a non-linear/polynomial relationship was observed (Schüldt and Harms-Ringdahl 1988; Solomonow et al. 1990). It has been suggested that the homogeneity of muscle fibres that make up the muscle is a contributor to the linearity or non-linearity of the force-EMG relationship (Guimaraes et al. 1994; Lexell et al. 1983). The homogeneity of muscle fibres making up muscles will be different which may explain the different relationships reported. The CES musculature is primarily made up of homogenous muscle fibres (Boyd-Clark et al. 2001) which is likely to be a reason for the linear relationship presented in this study. Both linear and non-linear models were fitted to the data and a linear model was found to provide the best correlations and therefore was the method chosen for the current study. The correlation coefficients were stronger than those describing polynomial trend, which was also fitted to these data during preliminary data analysis. The linearity of the correlation strengthened at forces <35N (Figure 4.10), which is consistent with the relationship described by previous researchers (Nigg and Herzog 2007; Solomonow et al. 1990). The strongest correlations were obtained for mean values of the right and left CES musculature; hence, the mean value is recommended when measuring EMG data for symmetrical tasks.

The mean (%) absolute error between determined and actual force in the force-EMG study was 5.80N (18.68%) (95% CI 4.84 – 6.76N; 95% CI 15.98 – 21.39%). The discrepancy between the two measurements may have come from a number of different sources relating to EMG measurement. Although participants performed the contractions for a short period of time, they would have started to suffer from the effects of fatigue during the randomised contraction trials. Muscular fatigue is known to negatively affect EMG signal (Disselhorst-Klug et al. 2009; De Luca 1997) causing a greater EMG amplitude than would have been measured had there not been any fatigue. Additionally, each participant was instructed to perform isometric contractions up to their MVC, to facilitate identification of a relationship between EMG and force. Adopting this technique of exerting up to MVC does, however, prove difficult to isolate specific muscles for EMG measurements. This means that participants may have inadvertently recruited surrounding muscles during contractions approaching MVC, to support CES contraction and, thus, aid and increase force production. This would have negatively affected the EMG signal, because only specific muscles were monitored and this would add muscular cross-talk into the signal, which is extremely difficult to quantify (Sommerich et al. 2000). Muscular cross-talk may be an explanation as to the frequent over-prediction of force through the use of EMG.

The results demonstrate a frequent over-prediction of force from EMG. This consistent direction of error enables a simple constant to be calculated and used to adjust the determined values. The absolute mean errors between determined and actual force were enhanced by the adjustment constant, therefore this adjustment is recommended for future application of such a method. The frequent over-prediction observed during the self-randomised contractions may be because these contractions had more dynamic properties than those used for the correlation trials. The mean absolute difference were less than 6N. This guides clinicians and sports scientists with understanding the degree of confidence with which EMG can be used to predict muscle force. The consistency of this protocol is evidenced by 90% of data points lying within the acceptable limit of error (Figure 32).

The results of this study suggest that the method outlined above could be used with confidence when the aim is quantifying the muscle force. This is especially important in a rehabilitation setting where gradual loading of damaged structures should form part of the rehabilitation protocols (Brody 2012; Adams and Dolan 2005; Setton and Chen 2006). Furthermore, it is known that paraspinal muscles alone can generate enough force to cause injury, including spinal fracture even in healthy spines (Melton et al. 2007; Mehlhorn et al. 2007; Youssef et al. 1976). Therefore, knowledge of muscle forces is important for rehabilitation, especially in regions of structural weakness following injury. The quantification of the effect of exercises and rehabilitation protocols on neck force will enable clinicians to make better choices as to appropriate exercises to be included and when. The method described above could be one way of further developing this knowledge and optimising the recovery of patients.

Individualised correlation curves were used in this study to predict muscular force production in recognition of the large variation in EMG amplitude that hinders the transferability of correlations between people (Queisser et al. 1994). This approach did, however, introduce inter-participant variation with, for example, differences in the range of contractions from different persons causing a clustering of data points at lower and higher ends of some force-EMG curves. Additionally, each participant was instructed to perform isometric contractions, to facilitate identification of a relationship. Adopting this technique does, however, prove difficult to isolate specific muscles for EMG measurements, meaning that the surrounding muscles may have been inadvertently recruited to support CES contraction, thus negatively affecting the EMG signal. Furthermore, the adoption of a position of 90º hip flexion may mean that back extensors are also used to aid force production. It is recognised that this method is for isometric and not dynamic contractions and thus will be more applicable to environments where there is little motion of the neck. It may prove a useful protocol however for sporting environments where direct force measurement is impossible and therefore an indirect measurement system is required. For example, a potential application could be during rugby scrummaging where there is a relatively small amount of motion of the neck and players also adopt a position of 90º hip flexion. EMG signals have a number of factors that can affect it and, as such, it is acknowledged that EMG-force correlation trials will have to be performed prior to each testing session without removing or changing the position of the electrodes. Carrying out a correlation trial prior to performing the action in question would provide a correlation that is specific to the conditions of that particular day and the participant.

7.3. Common Scrummaging Observations

Across the studies, it was possible to make a number of common observations such as the magnitude of muscle activity and the magnitude of thoracic and lumbar ROM utilised.

An important point to note with regards to normal spinal kinematics during different scrummaging scenarios is the principle of keeping ‘the spine in line’ which is a principle often coached at all levels of the game (Ieremia et al. 2002). It is apparent that the spine, particularly the lumbar spine, must be flexed in order to scrummage and thus a large amount of sagittal plane motion for this spinal region is required. It may be suggested, however, that motion in the transverse and coronal planes deviates from the ‘spine in line’ principle and thus, should be minimised as much as possible. Coupled motion will result in the off-centre application of compressive forces (Cazzola et al. 2014b) transmitted through the spine during scrummaging which may cause large bending moments (Dolan and Adams 2001) leading to buckling mechanisms, possible ligament damage and dislocations of the facet joints (Kuster et al. 2012; Dennison et al. 2012). Couple motion in each of the spinal segments can be observed in the dynamic ROM figures presented in this thesis. It can be observed that during some trials there is a peak in cervical motion in one plane and a concurrent peak in lateral bending or rotation for example. Since no significant differences ($p>0.05$) were observed between the two live scrummaging sequences (CBS vs CTPE), it may be concluded that the new laws do not significantly alter the segmental spinal kinematics of the playing position investigated.

Reducing chronic injury from the scrum environment is complex. Greater or excessive spinal motion is associated with chronic degeneration in all regions of the spine (Kumaresan et al. 1999; Adams and Hutton 1985b), thereby leading to conclusions regarding reduced ROM being beneficial for chronic injury potential. Indeed, only a modest range of cervical motion (~60% of total available range) was observed suggesting that excessive motion may not be the main source of degenerative change in the cervical spines of front row players. For a front row player to constrain motion in an oppressing environment requires significant muscle activity and this was shown in the final study of the thesis. This muscle activity equated, using the relationship developed, to ~230N of vertical force production across all conditions investigated. Moreover, this amount of load related to ~75% MVC (±20%) across conditions. Previous research has observed 60% MVC (±16%) and 62% MVC (±17) across two playing levels during individual machine scrummaging when investigating the same musculature (Piscione and Gamet 2006a). The relatively large standard deviations observed across these studies would indicate that there is no significant difference in CES muscle activity across the variety of conditions investigated. Additionally, there was slightly greater activity for one side of the CES muscles compared to the other. On the information sheet that the participants filled in, they provided information as to their dominant side; left or right. The side of the CES musculature that demonstrated higher activity correlated highly (R^2>0.8) with the participants' dominant side. The majority of players were right-side dominant and therefore it is interesting to note that the muscle activity was higher on the opposite side to where the ball was put in from.

This relatively high level of muscle activity would, of course, result in a cost of compressive load, also known to be linked to chronic degeneration (Skrzypiec et al. 2007). The relatively modest cervical ROM observed in this study is likely to be the result of a large amount of 'stabilising' muscle forces, where the demands of the dynamic competitive scrum see the hooker attempt to minimise head displacement. To this end, the muscle activity noted in previous studies (Cazzola et al. 2015; Piscione and Gamet 2006) and the current study would indicate relatively high compressive loads on the cervical spine (Skrzypiec et al. 2007). Unless this compressive load is outside of normal activity zones (i.e. greater than MVC), then injury is unlikely. Repeated exposure to sub-maximal loads, however, may lead to chronic changes. In addition to this, the cervical spine is known to be weaker in resisting compression in a flexed position compared to a neutral position. When in flexion, resistance to compression has been attributed to ~50% from the spinous processes and associated ligaments, ~30% from the zygapophyseal joints and ~20% from the intervertebral disc (Przybyla et al. 2007). Therefore, during scrummaging, when the cervical spine is in a position of flexion and there is compression from the second and back rows and the opposing team, injury to the above structures may be more likely.

This may provide some explanation as to the relationship between scrummaging and long-term degenerative changes in the cervical spine. Therefore, the hooker uses muscle activity to stiffen and prevent large displacements, resulting in the conservative cervical ROM values observed in this study; however, a large compressive cost is likely to burden the anatomical structures of the cervical spine.

The thoracic and lumbar spine display relatively large ROM during scrummaging. This is particularly evident for extension and lateral bending in the thoracic spine and flexion for the upper lumbar spine and side bending and rotation for the lower lumbar spine. For these motions, over 90% of the available range was utilised relative to the participant's maximum. This is significant as end-range spinal positions may result in reduced muscle activity (Olson et al. 2004), altered muscle function (McGill et al. 2000), increased load on the passive osteoligamentous spine (Goel et al. 1993) and increased risk of tissue damage (Chosa et al. 2004). Available motion in other planes, once the spine is positioned at end range, is also significantly compromised, perhaps owing to reduced tissue compliance and bony opposition (Burnett et al. 2008; Ebert et al. 2014); hence, the use of such large proportions of available range may leave the spine vulnerable to injury (Cholewicki and McGill 1996). Moreover, in vitro testing has shown that the lumbar spine is at a 50 fold increased risk of damage when in a flexed position compared to a neutral position when the same compressive load was applied (Gallagher et al. 2006).

With regards to the (determined) force data collected during this thesis, it is difficult to directly compare this with force data presented in previous studies. In previous machine scrummaging studies, the pusher arms of the scrum machine were instrumented to measure force in three dimensions (Milburn 1987; Preatoni et al. 2013; Quarrie and Wilson 2000) and a similar principle was used during live scrummaging (Cazzola, Preatoni, et al. 2015). This means that the vertical force reported is the vertical (z) component of the force vector created during engagement and sustained scrummaging. This is not the same as the vertical force reported in this study as it is the vertical component of force borne by the shoulders as opposed to the vertical force applied directly to the back of the head of the hooker. The vertical (z) component of the force vector reported in previous research ranged from -0.1 to 1.1kN and mean peak vertical force production in this study was 0.2kN. Thus, force magnitudes were within a comparable range although different variables were reported. As one of the aims of this study was to begin to understand cervical joint loading, it was the vertical force applied to the back of the head that was measured during machine scrummaging rather than the force vector when engaging with the machine or an opposing pack.

The data presented in the preliminary and engagement techniques studies demonstrated no difference in cervical sagittal ROM across the scrummaging conditions. The magnitude of spinal ROM utilised was largely variable depending on the specific participant. This large variability of values can be seen in the dynamic ROM traces (**Error! Reference source not found.** – **Error! Reference source not found.**). These figures further strengthen the notion that the new scrum laws neither positively nor negatively affect hooker spinal kinematics, owing to highly individual and variable ROM behaviour. This would suggest that spinal kinematics may not be readily influenced simply by changing scrum laws, as it would appear that intrinsic control of ROM is the main determinant of kinematics during scrummaging. Furthermore, participant's intrinsic control of their ROM may be indicated by the coefficient of variation data presented (Table 63 - Table 66). Lower CoV values suggest that the participant had a better control of their ROM for that particular segment and motion although there was no pattern to suggest that variation for a specific segment and motion was consistent across participants. Thus, it may be suggested that each participant has different physical characteristics that give them different intrinsic control of their ROM during scrummaging and must adapt and react to the specific requirements of the scrum.

This finding that scrummaging is highly individual and variable may be significant from a coaching perspective. Standard coaching protocols may not be hugely effective as the nature of the scrum is so difficult to predict. Therefore, once an appropriate level of experience is gained, it may be most beneficial for the player to take part in live scrummaging as it has been advised that this is the best scenario in which to develop specific neuromuscular activation patterns (Cazzola et al. 2015).

Although there is evidence to suggest that loading the spine towards the end of its range is a significant risk of injury, as previously mentioned, there is also evidence to suggest that more constrained kinematic conditions may lead to repeated stresses on the same vertebral structures (Adams & Hutton, 1985b; Adams, McNally, Chinn, & Dolan, 1994; Adams, McNally, & Dolan, 1996) although these studies were performed in vitro and on specimens of an ageing population. Therefore, the link between more constrained kinematic conditions (e.g. machine scrummaging) and reduced injury risk is not quite as straight forward as it may initially seem. For example, flexion reduces stress in the apophyseal joints and posterior half of the annulus fibrosus but it also increases stress on the anterior annulus (Adams & Hutton, 1985b). Thus, the suggestion that reduced ROM magnitudes may reduce injury risk must be interpreted with caution owing to the aforementioned reasons.

7.4. Preliminary In-Field Testing

The preliminary ‘in-field’ study was the first of its kind to evaluate spinal kinematics in detail of one particular playing position. No previous scrummaging studies (Sayers et al. 2009; Preatoni et al. 2015; Cazzola et al. 2015) have investigated spinal kinematics to this level of detail. From this study, two problems were identified with regards to the placement and orientation of the inertial sensors. During preliminary testing, the inertial sensors were placed with their longest dimension running laterally across the torso. This resulted in some difficulty adhering the sensors to the participant at the specific landmarks owing to the large musculature of rugby players and therefore resulted in the loss of some data as sensors would fall off during experimentation. With the sensors in this orientation, some of the footprint of each sensor was not in contact with the skin of the participants resulting in poor skin adhesion. It was therefore decided that the sensors should be rotated anticlockwise by 90° for future studies so that a greater part of the sensors’ footprint was in contact with the skin reducing the likelihood of sensors falling off during the trials. As well as the orientation, the placement of the sensors along the spine was changed. In this study, the thoracic spine was divided into three segments which allowed for the best spacing of the sensors down the spine. This was changed, however, as segmenting the lumbar spine would provide greater kinematic detail of this region and the lumbar spine is at risk of injury during scrummaging

(Fuller, Brooks and Kemp 2007). Therefore, the sensors were moved so that two thoracic and two lumbar segments were investigated. For the subsequent studies, the spinal landmarks used were C7, T7, T12, L3 and S1.

With regards to the force-EMG data, it was not possible to identify a relationship between force and EMG with the EMG data that was obtained. In the laboratory study, the datasets were synchronised by maximal EMG and maximal force which were easily identifiable. The aim during the in-field study was to similarly identify maximal muscle activity and force and then work backwards to create an individualised force-EMG correlation curve. This, however, was not possible as the maximal EMG activity was unidentifiable making it impossible to create a correlation curve for each participant. This problem was addressed for future studies by changing the EMG equipment so that a robust correlation curve could be obtained and therefore force could be determined.

It is acknowledged that only a limited number of participants were recruited to take part in this study. This study was used as a pilot to find and address any problems with the methods used.

7.5. Playing Surfaces Study

Hooker peak ROM during the scrum did not change significantly with respect to the playing surface; however, the synthetic surface produced a more conservative ROM in nearly 80% of the 30 variables. The natural surface produced greater angular velocities in right lateral bending and left rotation, and left lateral bending, in the lower thoracic segment. Although these differences were not statistically significant after Bonferroni correction, the size of the effect was large ($d>0.8$). In addition to this, the standard deviations of peak motion on 3G were lower than that of grass. These lower values would indicate greater postural control and stability as players are making less adjustments.

Whilst no injury data was collected during this study, the results suggest that there may be a trend towards a slightly reduced injury risk when scrummaging on synthetic surfaces owing to greater stability. Spinal angular velocity has a direct relationship to trunk muscle activity with increased angular velocities resulting in much increased paraspinal muscle activity (Fan, Liu, & Ni, 2014; Mawston & Boocock, 2012; Williams, Haq, & Lee, 2013b). Increased paraspinal muscle activity has been shown to cause increased spinal compression in the cervical (Skrzypiec et al. 2007), thoracic (Caneiro et al. 2010) and lumbar (Adams & Hutton, 1982, 1985a) regions. The spine is always under compression but greater muscle activity will increase the magnitude of compression and therefore, the magnitude of load being borne by the vertebrae. It remains unclear at this stage as to whether the muscles react to the externally driven rotations or whether they are the cause of the greater angular velocities.

In the lumbar region, increased compressive loading has been shown to be a high risk of endplate fracture and, when combined with bending, may cause injury to the intervertebral disc (Adams & Hutton, 1982, 1985a). It must be noted that the spine is always in compression and therefore the loads used in vitro were used to simulate loads over and above normal physiological loading. Furthermore, there is a relatively high prevalence of thoracic spine injuries reported in rugby forwards from T8-T12 (Hind et al. 2014). In the cervical spine, compression causes the loss of intervertebral disc height and resultant increased load bearing on the neural arch and uncovertebral joints (Skrzypiec et al. 2007). Over a prolonged time, this may lead to the development of degenerative changes in the spine such as the formation of osteophytes (Kumaresan et al. 2001). These degenerative changes have been observed previously in front row players in the cervical spine (Scher 1990b; Berge et al. 1999). Whilst it is not known whether similar pathologies occur in the thoracic spine, it appears likely, that owing to the responses shown to similar increases in muscle activity for other regions of the spine, similarity can be determined for the effects on the thoracic spine.

The cervical spine is widely reported in the literature to suffer from the greatest number of injuries during scrummaging (Quarrie et al. 2002; Wetzler et al. 1998; Scher 1982) and front row players are well documented to suffer from premature chronic degeneration of the cervical vertebrae (Scher 1990b; Broughton 1993; O'Brien 1996; Berge et al. 1999). The data presented relating to angular velocities (Table 36) of the cervical spine is relevant to this as it may provide some information as to why these chronic injuries are so prevalent.

From the data, it can be seen that mean peak cervical spine angular velocity was greater on natural turf than synthetic turf. For flexion, left lateral bending and right rotation, there was a medium effect size (d>0.5). Greater angular velocities means greater loading/force of the structure in question (Yoganandan, Pintar, Sances Jr, Reinartz, & Larson, 1991; Yoganandan & Pintar, 1997; Yoganandan, Pintar, Cusick, & Hollowell, 1999) because of the increased muscle activity of paraspinal musculature. The repetitive loading experienced during scrummaging (Scher 1990b), with forces, (Nightingale, Richardson, & Myers, 1997; Nightingale et al., 1997; Yoganandan et al., 1991) velocities and accelerations (Portero, Quaine, Cahouet, Thoumie, & Portero, 2013; Yoganandan et al., 1991) that are comparable to the current data and other published data on scrummaging (Cazzola et al., 2014; Milburn, 1993; Preatoni et al., 2013; Preatoni, Stokes, England, & Trewartha, 2014; Quarrie & Wilson, 2000) may, with time, lead to chronic degenerative changes and neck pain (Lark and McCarthy 2010; Pinsault et al. 2010; Scher 1990b; Berge et al. 1999). Thus, a reduction in angular velocity is likely to be a positive outcome for the playing position considered as it may delay the onset of chronic degenerative changes that are often seen in these players.

Scrum stability is extremely important to try and reduce the number of collapses and therefore, reduce the risk of catastrophic spinal injury. Stability was estimated by considering the magnitude of kinematic variables where lower magnitudes were taken to mean more stability for the player being investigated. This approach is similar to that adopted by Cazzola et al (2014), where lower excursions/ROM were considered to mean greater stability, since players made less postural adjustments. There has been some anecdotal evidence to suggest that scrums are more stable on synthetic surfaces, as there was an observed decrease in the number of collapsed scrums (BBC 2013), but this is the first study to provide empirical evidence to suggest that scrummaging on a synthetic surface does have a potentially positive effect on stability for the player that was investigated. Moreover, when looking back at the level of rainfall during data collection, South Wales experienced a relatively high amount of rain compared to other regions of the UK. This would indicate that the grass surfaces would have had poor grip and therefore, may provide an explanation for the greater kinematic magnitudes observed on this surface.

7.6. Engagement Techniques Study

This study set out to determine whether the evolution of rugby laws relating to the scrum had any effect on hooker spinal biomechanics during live scrums. The results presented here indicate that the change in laws do not have any significant effect on hooker spinal biomechanics when comparing between the CBS (new), and CTPE (old), scrummaging technique. There were some kinematic differences between machine and live scrummaging trials but only for the upper spinal regions. This study performed the first multi-regional spinal analysis, generating data that enhances our understanding of spinal kinematics within both a machine and live environment, and using the new and old scrummaging techniques.

Previous studies have demonstrated that the move to CBS engagement has significantly altered the scrummaging biomechanics (Cazzola, Preatoni, et al. 2015; Cazzola et al. 2014a). Indeed there is growing evidence that the impact of engagement is reduced using CBS as measured by the impact to the shoulder girdle, both in machine (Preatoni et al. 2015) and live (Cazzola et al. 2014b; Cazzola et al. 2014a) scrummaging. This was mainly owing to the reduced distance between front rows prior to the impact phase. The current study did not measure the impact phase specifically or impact at the shoulders. Instead, it sought to further our understanding of the law change on other biomechanical aspects. To this end, the findings of the current study do not support or refute the recommendations to move to the new laws. A synthesis of results would suggest that the new laws result in reduced shoulder impact and have no effect on spinal kinematics.

The data presented in this study from 29 hookers demonstrated no difference in spinal ROM across the scrummaging techniques. The magnitude of spinal motion utilised was quite varied and was largely dependent on the specific participant and the nature of the scrum. This large variability of values can be seen in the dynamic ROM traces during scrummaging (Figures 47-56). These figures suggest that spinal kinematics may not be readily influenced simply by changing scrum laws, as it would appear that intrinsic control of ROM is the main determinant of kinematics during scrummaging. Furthermore, it is noted that only 63% of total sagittal ROM was used regardless of engagement sequence suggesting that both sequences (CBS and CTPE) represent a low risk of hyperflexion for most participants, a mechanism believed to be linked to catastrophic injury (Kuster et al.

2012). It may be that catastrophic hyperflexion injuries are the sole domain of the collapsed scrum; however, a more substantive claim can only be levied on observing a greater number of scrums.

The data also indicated that machine scrummaging utilised significantly less ROM for the upper spinal regions. This may be owing to the restricting physical environment of the machine or that these non-competitive scrums demand less motion. Previous studies have demonstrated less cervical muscle demand of machine versus live scrummaging (Cazzola et al. 2015). Therefore the body of evidence seems to suggest that machine scrummaging requires significantly less cervical and upper thoracic ROM and less cervical muscle activity and thus, is an environment which poorly reflects the true cervical demands of live scrummaging. This may be important from a rehabilitation perspective where lower demands might be desirable, say after cervical injury, however it should not be used to replace the demands of live scrummaging.

When considering live scrummaging and playing surface, there was some interaction between these independent variables for peak ROM. This was only evident for left lateral bending of the cervical spine. On the 3G surface, there was a reduction of 10° going from CBS to CTPE and on grass, there was a slight increase of 2° from CBS to CTPE. Whether such changes in ROM result in clinically important differences in such a variable environment is yet to be established.

In performing the first multi-regional spinal kinematics evaluation of the hooker *within* the rugby union scrum, this study is unique in presenting the greatest detail of spinal kinematic data. This analysis revealed that, when considering the relative motion of the 5 spinal regions, there was insignificant b variation between the new (CBS) and old (CTPE) techniques. Hence, whilst the objective of reducing chronic spinal injury appears to remain unmet, this study adds to the debate by using novel methods to report the first spinal kinematics from within a competitive environment.

7.7. Scrum Collapse

During data collection, three scrum collapses occurred. This is unlikely to be representative of the number that occur during competitive matches (Quarrie 2009). Whilst this is an extremely low number across all the trials performed, it does provide an insight into the biomechanics of an event that has not been characterised. It is known that no meaningful conclusions can be drawn from such a small sample size. It is possible, however, that these events are unique to some extent and therefore, even a large number of observations would make it difficult to draw any meaningful conclusions from. Scrum collapses have been documented to cause some of the most severe injuries during rugby scrummaging (Broughton 2009; Secin et al. 1999; Scher 1982) and are also associated with a much increased risk of injury (Taylor et al. 2014; Roberts et al. 2014). Additionally, a number of authors have suggested hyperflexion as the mechanism of injury (Sovio et al. 1984; Scher 1982) as well as with concurrent compression and/or rotation (Scher 1990a; Scher 1982). From the results presented in this thesis, the cervical segment is forcibly subjected to flexion primarily, with large magnitudes of concurrent lateral bending and rotation.

During all three collapses, the individual experienced large magnitudes of coupled motion but never exceeded their voluntary, active ROM. This would suggest that an injury is unlikely and, indeed, none of these players were injured. Coupled with the driving force of the second and back rows as the player was driven into the turf, however, may put these players at a much increased risk of injury as some form of vertebral failure is likely when the spine is in non-neutral positions (Przybyla et al. 2007; Aultman et al. 2004). This is because the absolute magnitude of load the spine can withstand in a particular direction is less when the spine is in a non-neutral position. This sort of mechanism has been shown, in vitro, to result in blunting or wedging of the vertebral body to severe fracture dislocations (Berry and Rao 2013), fracture of the C1 vertebral body (Nightingale et al. 1996), and bilateral facet dislocation (Bauze and Ardran 1978). Exact forces the spine is subjected to during scrummaging remain unknown, therefore, the link between in vitro testing and possible spinal injury mechanisms during scrummaging are, to some extent, speculative. Acute spinal injuries recorded resulting from scrum collapse have been fracture dislocation of the C4/5 or C5/6 junction (Quarrie et al. 2002), unilateral facet dislocation (Dennison et al. 2012) or facet locking (Scher 1982) in front row players. These types of injuries are more likely to occur extrapolating from our knowledge of recorded injuries to front row players and in vitro testing.

As previously mentioned, none of these players were injured, however. To the author's knowledge, in vitro testing is the only source of information to which comparisons can be drawn of these biomechanical findings. Furthermore, the contribution of soft tissue in dynamic loading environments is likely to significantly alter the mechanical response of the spine (Kumaresan et al. 1999; Yoganandan et al. 2001). This contribution of soft tissue is extremely difficult to replicate during in vitro laboratory testing and, as previously mentioned, significantly alters the mechanical response of the spine (Kumaresan et al. 1999; Yoganandan et al. 2001). Furthermore, front row rugby players in particular are likely to exhibit physical and musculoskeletal characteristics that are not represented in the general population (Fuller et al. 2013).

Although performed in the most game-like environment, competitive training, it was found very few scrums collapsed. Research has shown that in competitive matches, over 12% of scrums collapsed although this evidence is now dated (Quarrie 2009). Despite the law change, it is unlikely that only 3 scrums in over 200 total scrums (both CBS & CTPE) investigated would collapse. If the above statistic carries over to the new laws, at least 25 scrums would have collapsed. This suggests that players may not scrummage as physically or competitively against team mates as they do during a competitive match.

7.8. Strengths and Limitations

This thesis was the first set of studies to report detailed spinal biomechanics successfully combining both kinematic and muscle activity/determined force data. It was the first of its kind to attempt to relate EMG to force and use this relationship during live scrummaging to try and begin to quantify cervical spinal loading.

The study was designed to try and minimise any potential for experimental bias. The participants were randomly allocated to groups indicating the order in which they would perform the live scrummaging trials to minimise any allocation bias. Participants were blinded to the hypotheses of the trials to minimise bias they might have for either engagement sequence. The participants were, however, informed of the two different conditions in advance so as to prepare themselves for each sequence. The instructions were standardised by the provision of participant details forms so all participants had the same instructions whilst the effects of fatigue were minimised by allowing the participants two minutes to allow for physiological recovery (Netto et al. 2007; Burnett et al. 2007). This was particularly important for monitoring muscle activity since fatigue can have a significant effect on muscle activity amplitude (Basmaijan 1978; Disselhorst-Klug et al. 2009). Furthermore, as outlined in the method, as many variables as possible were kept constant such as the hooker in question always being part of the attacking scrum and therefore the ball was always fed in on the same side and the same scrum half was used to feed the ball in to minimise the effect of 'angle of put in' on hooker spinal motion. Some scrum halves may angle the ball towards their team slightly more than others meaning the 'hook' may be less pronounced and therefore lumbar rotation is likely to be less.

One particular strength of this thesis is the level of detail for the kinematic data collected. In the most recent studies performed in both machine (Preatoni et al. 2013; Preatoni et al. 2015) and live (Cazzola et al. 2015) scrummaging, high speed (200Hz) and normal speed (50Hz) video were used to assess scrummaging kinematics of each individual player of the scrum. The high speed video was placed above the scrum with points on both shoulders and the sacrum marked (Preatoni, Wallbaum, et al. 2012). This allowed the direction and angle of the trunk, as a whole, to be monitored during engagement and sustained scrummaging. The problem with this, however, is that the trunk was modelled as one rigid segment and no detailed data of segmental spinal motion could be obtained. Furthermore, only two-dimensional data could be obtained from this kinematic analysis. It did, however, mean that trunk angles of all the players were considered unlike the present study where only one player was considered. Previous to this, Sayers (2011) performed a kinematic analysis of high-performance prop forwards. This particular study focussed on two dimensional analysis of the full body. Segments were defined as the ankle, shank, thigh, trunk, head and arms. This, in some respects, is more detailed than the current study as it provides data of multiple body segments rather than just segmenting the spine. The problem is, however, that the analysis was only two dimensional and therefore one plane of motion was completely neglected for every segment. Furthermore, the study also analysed 3-, 5-, and 8-man scrums. The lower number of players in the scrum did not accurately represent

8-man scrum kinematics and therefore this study used solely 8-man scrums to investigate spinal kinematic parameters as this is the most accurate platform to investigate hooker spinal kinematics.

A limitation of the playing surfaces study was that the groups were not the same on both surfaces. Ideally, one group of players would have taken part in trials on both playing surfaces which would mean a more robust statistical test could have been used. Owing to the availability of pitches to different teams, it was not possible to do this and this limitation is acknowledged by the author. The groups, however, did not differ statistically ($p>0.05$) in terms of their anthropometrics, ROM and background information collected which would suggest similarity between the groups. It may have been preferable, however, to individually match each player with a similar player from the other group.

A useful extension to the work presented would have been to annotate a “typical” dynamic ROM trace with the events that occurred during the scrum. Although this would have been a useful way in which to present the data, it was not possible as these points within the scrum were not easy to identify during data collection. Furthermore, the data demonstrated a large amount of variability across all studies which has been presented in all three experimental chapters. This made it very difficult to identify what could be characterised as a “typical” trace. The magnitude of spinal motion utilised was quite varied and was largely dependent on the specific participant and the nature of the scrum. This large variability of values can be seen in the tables of coefficient of variation values previously presented. These figures suggest that it is difficult to identify a “typical” trace, as it would appear that intrinsic control of ROM is the main determinant of kinematics during scrummaging.

A limitation of the thesis’ investigations is that it was conducted during training, which does not necessarily represent the biomechanical loading patterns of the more aggressive and physical nature of a competitive match against a rival team. It is, however, currently the best available scenario in which this unique biomechanical scenario can currently be investigated.

Another limitation that needs to be acknowledged is that the data presented in this thesis are compared to in vitro and computational modelling studies and therefore, the conclusions that can be drawn are speculative in nature. This is because our general knowledge of injury mechanism literature is being transferred to a specific biomechanical scenario. To the author's knowledge, currently, in vitro and computational modelling studies are the only available basis of reference to compare this data to. Furthermore, the rugby scrum is a unique biomechanical environment and its biomechanical features are not likely to be well represented in the literature compared to more common activities like that of in vitro representations of car crash biomechanics (Newman et al. 2000; Mertz and Patrick 1971; Enouen 1986). Furthermore, it is likely that rugby players may possess characteristics of postural control and general musculoskeletal characteristics that are not well represented in the average sample of subjects used in typical studies although this was not tested during the studies of the thesis. One such difference would be the age of rugby players compared with the age of specimens used as a comparison from biomechanical literature. This would mean transferring our knowledge of in vitro mechanisms may not be directly applicable to the rugby scrum scenario in which biomechanical loading is unique but is currently the best comparison that can be made.

A problem with regards to the use of inertial sensors is soft tissue motion artefact. The movement of soft tissue is what is recorded by the inertial sensors and not the movement of the spinal landmark. To the author's knowledge, the magnitude of soft tissue artefact has not been quantified for the different regions of the spine. Soft tissue artefact is a particular problem when the landmark is question has a large amount of soft tissue overlying it. An example of this would be the femur. If a sensor is placed on the femur, as the quadriceps contract, the orientation of the sensor will change when no actual motion will occur. Compared to this example, the spinal landmarks chosen have little underlying soft tissue and therefore the magnitude of these errors will be less. Although this is an error that needs to be acknowledged, the only way to overcome it would be to use invasive techniques so that the centre of rotation is at the landmark in question. This problem of skin motion artefact affects many motion analysis systems where markers are attached to the skin such as opto-electronic and electromagnetic tracking systems. For these techniques, this limitation must be acknowledged.

The relationship developed between force and EMG provided the opportunity to begin to quantify cervical spine loading during scrummaging for the hooker. The muscle activity being measured relates only to the resistance of the CES musculature to load applied to the back of the head. Without the musculature being active, any load in this direction would result in flexion of the cervical segment. This relationship, however, did not account for loads applied in any other direction. Therefore, any axial or anterior loading would not have been accounted for and the musculature being measured would not have provided a response to these loads. Therefore, any loading in any direction other than applied downwards to the back of the head remain unknown as is activity from other muscles.

Despite the limitations of the studies performed, this thesis provides valuable data relating to spinal biomechanics of the hooker during rugby scrummaging and how the evolution of rugby has influenced the potential risk of spinal injury. It has also provided a novel method in order to determine spinal biomechanics and it is hoped that these methods will be useful for further study both within the current field and other fields of biomechanics research.

8. Research Conclusions and Further Work

The main aim of this thesis was to initially develop a method to analyse spinal biomechanics during rugby scrummaging and then use this method to answer some specific research questions. The conclusions of the thesis are provided here in light of the aims of each part of the research.

8.1. Inertial Sensor Validation

The aim of this study was to demonstrate the validity of the chosen inertial sensors (3AMG, ThetaMetrix, Waterlooville, UK) against a known output of orientation for all three axes. Inertial sensors have previously been used in biomechanics research and therefore the aim was to determine the validity of these specific sensors for use during spinal kinematic analysis of rugby scrummaging. All axes of all 6 sensors demonstrated extremely high coefficients of concordance (>0.98) against the digital encoders of the rotary table showing that these sensors are a reliable measurement of orientation.

8.2. Force-EMG Study

The aim of this study was to determine whether EMG alone can be used as a predictor of cervical erector spinae (CES) muscular force production using individualised correlation curves. It was found that if individualised correlation curves of force against EMG were used, it was possible to use EMG alone to predict CES muscular force production. Determined force and actual force were not statistically different (p>0.05) and an enhanced p-value was determined when a simple adjustment coefficient was used to adjust determined force values. This was done as EMG often over determined force and therefore, a simple coefficient could be calculated to adjust these values.

8.3. Preliminary In-Field Testing

The aim of this study was determine whether the methods developed within the laboratory were applicable during in-field testing and also address any problems that were encountered during data collection. It was determined that segmental kinematic analysis of the spine was possible but problems with collection of the force-EMG data needed to be addressed. These problems were addressed for the final study of the thesis. The kinematic data showed that there were no statistically significant differences ($p<0.05$) in peak ROM and angular velocity between machine and live scrummaging. Furthermore, the variability (coefficient of variation) of peak ROM data for machine and live scrummaging was relatively large indicating that participants must adapt and react to each individual scrum. Thus, specific tactics for the hooker within the scrum may be difficult to implement as the scrum seems to be a situation in which the hooker must react to the situation presented at that time.

8.4. Effect of Playing Surface on Hooker Spinal Kinematics

This study aimed to determine whether there was any effect on hooker scrummaging spinal kinematics when changing playing surface from grass to 3G. Although no significant differences ($p>0.05$) were observed for peak ROM and peak angular velocity data for all segments, the magnitude of the effect was large ($d>0.8$) for some variables. There was a decrease in peak lower thoracic angular velocity in left and right lateral bending and left rotation when moving from grass to 3G. These reductions suggest that there may be slightly more spinal stability when scrummaging on a 3G surface compared to grass as players use lower ROM excursions and therefore undertake fewer postural adjustments. Previous research has not been able to definitively identify differences in the mechanical properties of grass and 3G turf and it is suggested that it is the consistency in the property of 3G turf in a variety of weather conditions that give rise to this increased spinal stability of the hooker.

8.5. Effect of Engagement Technique on Hooker Spinal Biomechanics

This study aimed to determine whether any differences were present for hooker spinal biomechanics during three scrummaging conditions. These conditions were machine scrummaging and live scrummaging of old (CTPE) and new (CBS) engagement sequences. With regards to live scrummaging, the study was unable to provide stand-alone evidence that the new engagement laws positively or negatively affect hooker spinal biomechanics for the variables investigated. That is, no significant differences ($p>0.05$) were observed for peak spinal kinematics of each segment and for peak muscle activity/determined force data. The insignificant change in muscle activity between live scrummaging techniques suggests that chronic cervical spine degeneration as a result of paraspinal muscle activity may be unchanged. It is difficult to say this with certainty, however, without long-term epidemiological studies evaluating the effects of the change in scrum laws.

Machine scrummaging consistently produced more conservative magnitudes of kinematic data and therefore, it is suggested that this training method is unlikely to accurately represent the complex biomechanical loading scenario of the live scrum. It is, however, a useful tool that can be used to practise ‘ideal’ scrummaging technique. Machine scrummaging and live scrummaging of both engagement sequences did not differ significantly ($p>0.05$) when considering peak cervical erector spinae muscle activity.

8.6. Directions for Future Work

The primary aim of this research was to establish biomechanical parameters of the spine of the rugby union hooker in a number different scenarios and see how these changes affect this player’s spinal biomechanics and potential injury risk. The methods may, however, be used in a variety of other scenarios and potential applications and directions for future work are explored here.

8.6.1. Application of Method in Other Research Areas

Initially, a method was developed to evaluate spinal kinematics and cervical spine loading which was used to asses spinal biomechanics during rugby scrummaging. This method, however, may have applications in other areas of sports biomechanics research. One such area could be in weight/power-lifting. This is particularly the case for spinal kinematics. Inertial sensors may provide valuable data relating to spinal motion and, as an extension to the method, a degree of curvature could also be measured. This could be used to compare how lifters technique may differ for the same type of lift and this method could be used to characterise what is considered as ‘good’ and ‘bad’ technique. The force-EMG method may also provide data relating to loading of the cervical spine in a dynamic activity where neck motion is minimal. This data may indicate whether the cervical spine musculature stiffens prior to performing a lift such as the deadlift or clean. This may be a neuromuscular activation strategy that aids the spine in terms of stability prior to the load being lifted.

8.6.2. Future Scrummaging Research

The methods used in this study could be used to assess biomechanical differences of hookers of different playing levels. This may provide an insight into how professional hookers scrummage from a biomechanical perspective and this data may help coaches teach young players optimal scrummaging technique in order to increase performance and reduce injury risk. Furthermore, this thesis concentrated on male, adult, amateur-level hookers who are likely to represent the vast majority of the rugby playing population. High-performance players are likely to exhibit physical and anthropometric characteristics not present in the majority of the rugby playing population and therefore may demonstrate a different scrummaging technique.

The method may also be used to assess both the loose-head and tight-head props. It is likely that the biomechanical loading experienced by these two players is very different to that of the hooker as they transmit a large amount of forward force.

Furthermore, this thesis concentrated solely on quantifying peak kinematic parameters but it may be possible, however, to provide dynamic postural/curvature analysis during rugby scrummaging. This may provide data as to how spinal posture changes during specific scrummaging events such as the impact and sustained scrummaging phase. It may also determine whether there are any pre-cursors to scrum collapse. For example, it may be evident from postural analysis that the lumbar spine is affected just prior to collapse and is a pre-cursor to the scrum collapsing.

Finally, it is essential to build on the data presented here and our current knowledge of spinal injury biomechanics in the rugby scrum in order to try and gain a better understanding of the causes of spinal degeneration and acute injuries. The understanding of spinal injury mechanisms in the scrum is vital in aiding the prevention of such injuries. Understanding injury mechanisms may also help coaches educate players from a young age about the risks and potential hazards of scrummaging whilst also providing effective scrummaging technique training to minimise the potential risk to these injuries.

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Appendix A:

Thesis Appendices

A.1. Additional Data

A.1.1. Inertial Sensor Validation Data (Sensors 2-6)

A.1.1.1. Roll Data

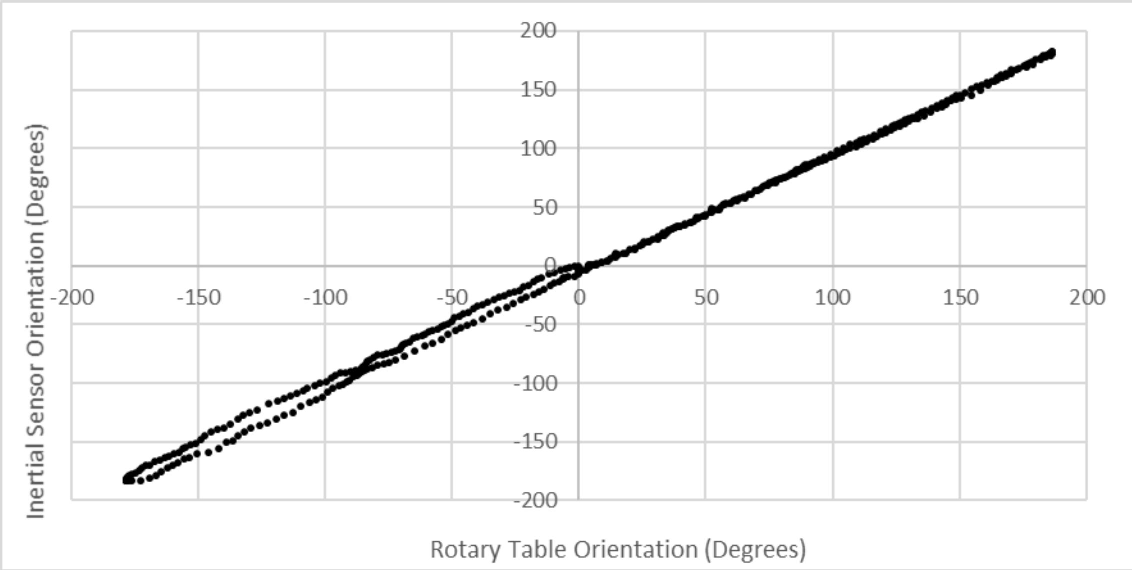


Figure A.1 - Rotary table orientation (x-axis) against inertial sensor 2 orientation (y-axis) for roll

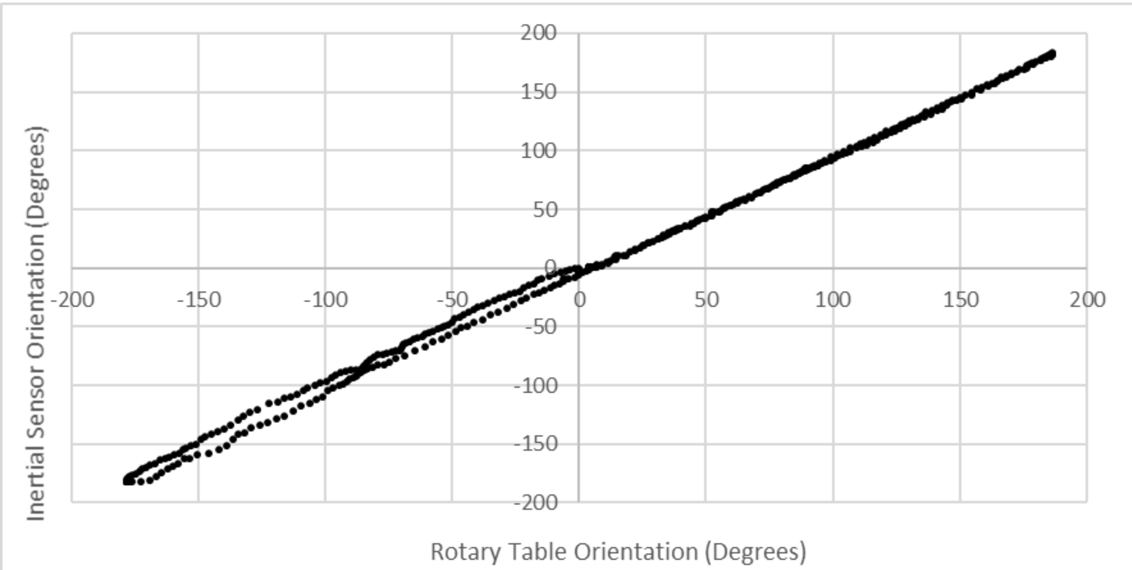


Figure A.2 - Rotary table orientation (x-axis) against inertial sensor 3 orientation (y-axis) for roll

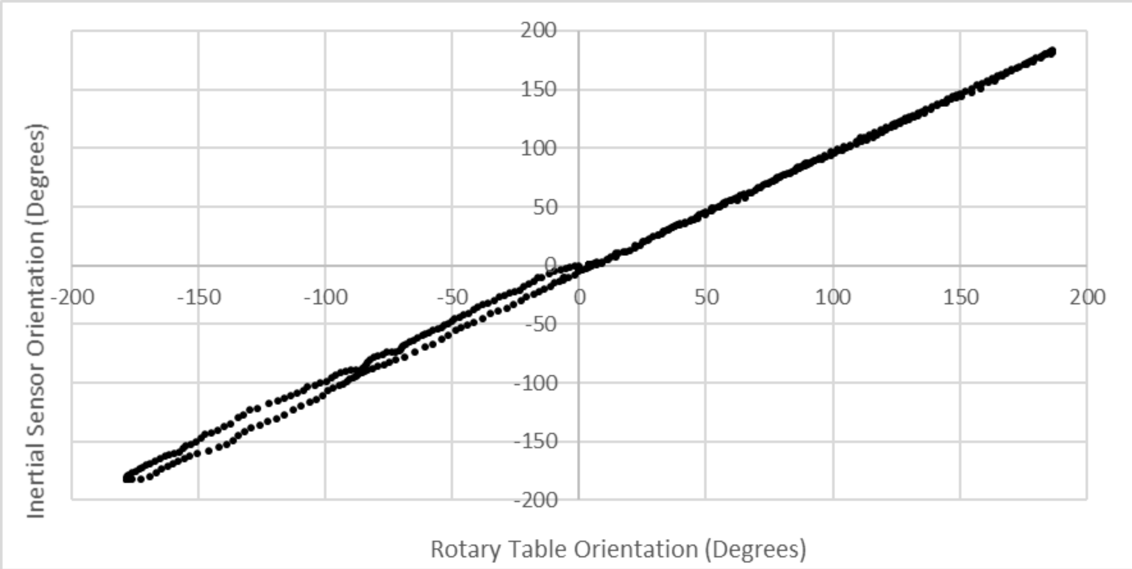


Figure A.3 - Rotary table orientation (x-axis) against inertial sensor 4 orientation (y-axis) for roll

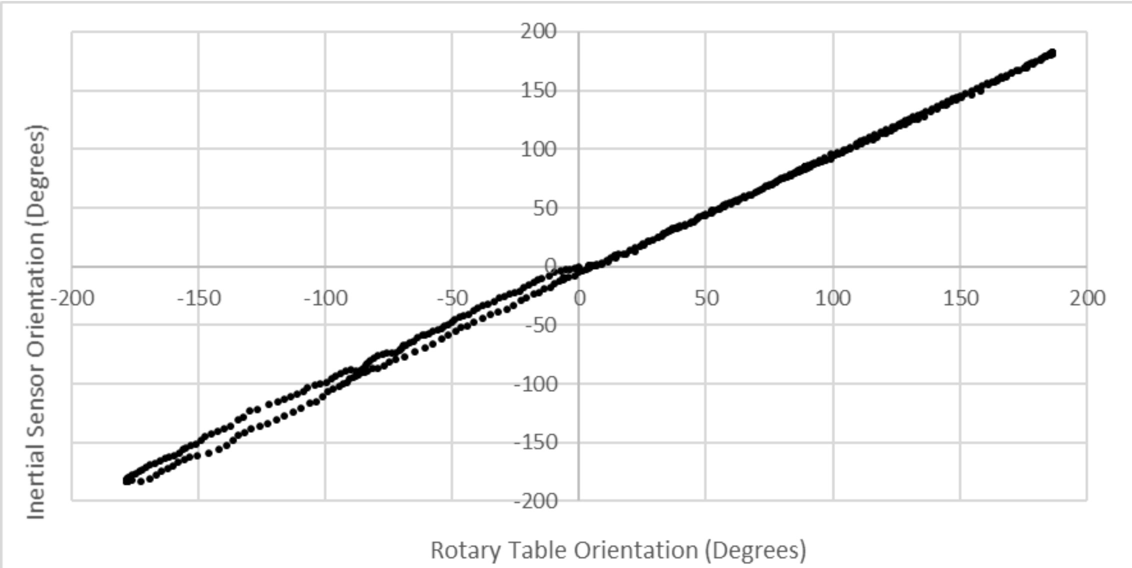


Figure A.4 - Rotary table orientation (x-axis) against inertial sensor 5 orientation (y-axis) for roll

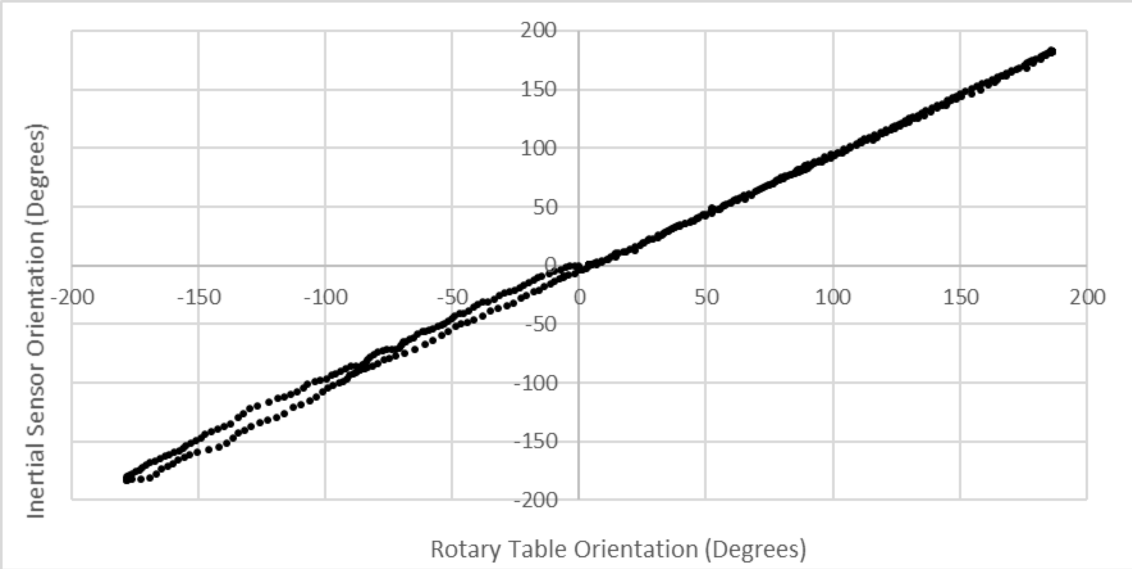


Figure A.5 - Rotary table orientation (x-axis) against inertial sensor 6 orientation (y-axis) for roll

A.1.1.2. Pitch Data

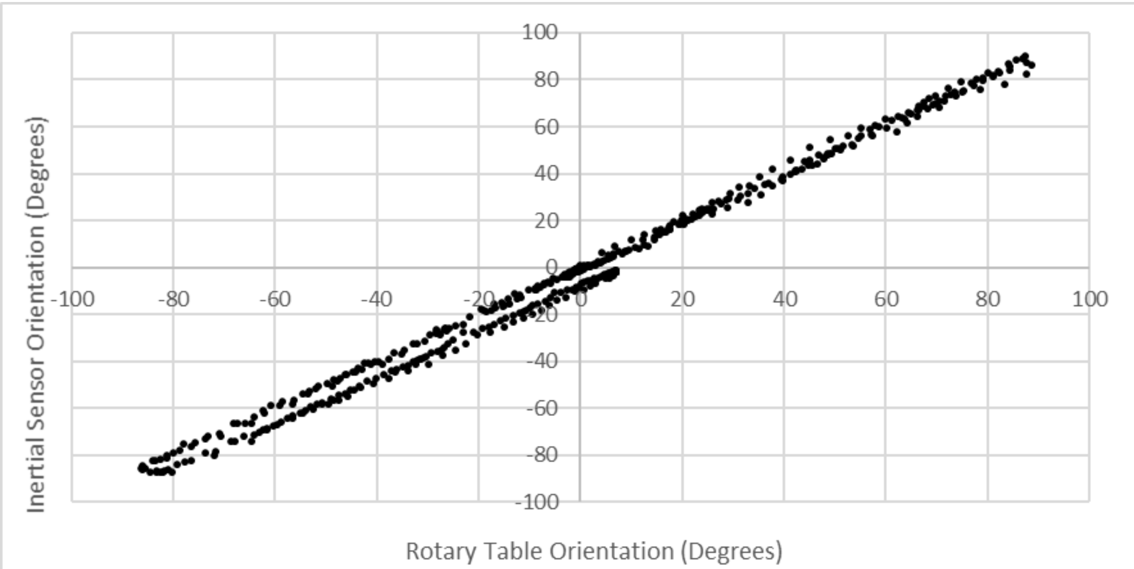


Figure A.6 - Rotary table orientation (x-axis) against inertial sensor 2 orientation (y-axis) for pitch

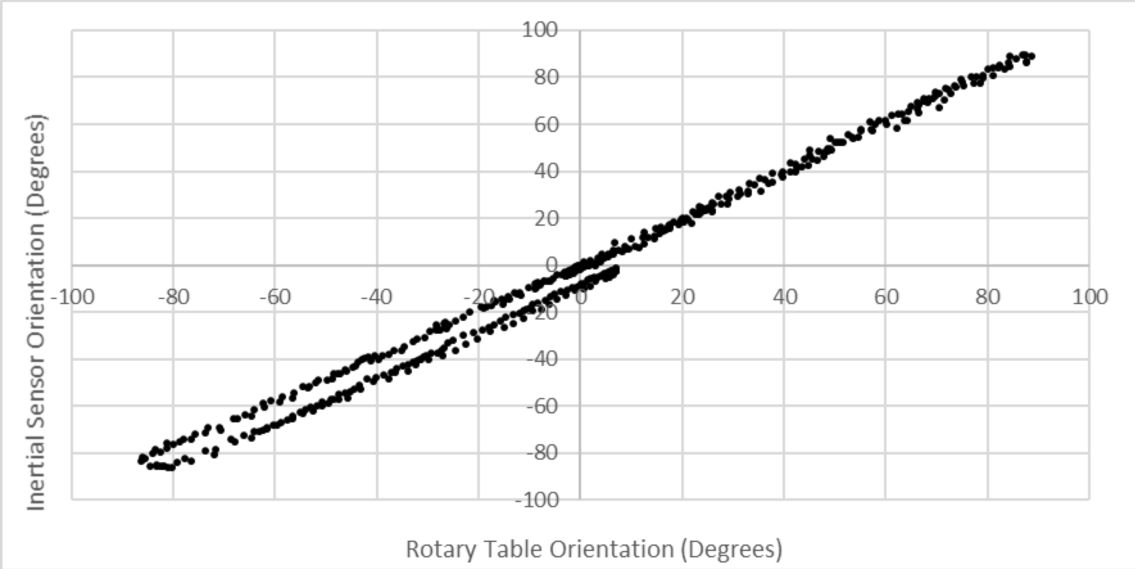


Figure A.7 - Rotary table orientation (x-axis) against inertial sensor 3 orientation (y-axis) for pitch

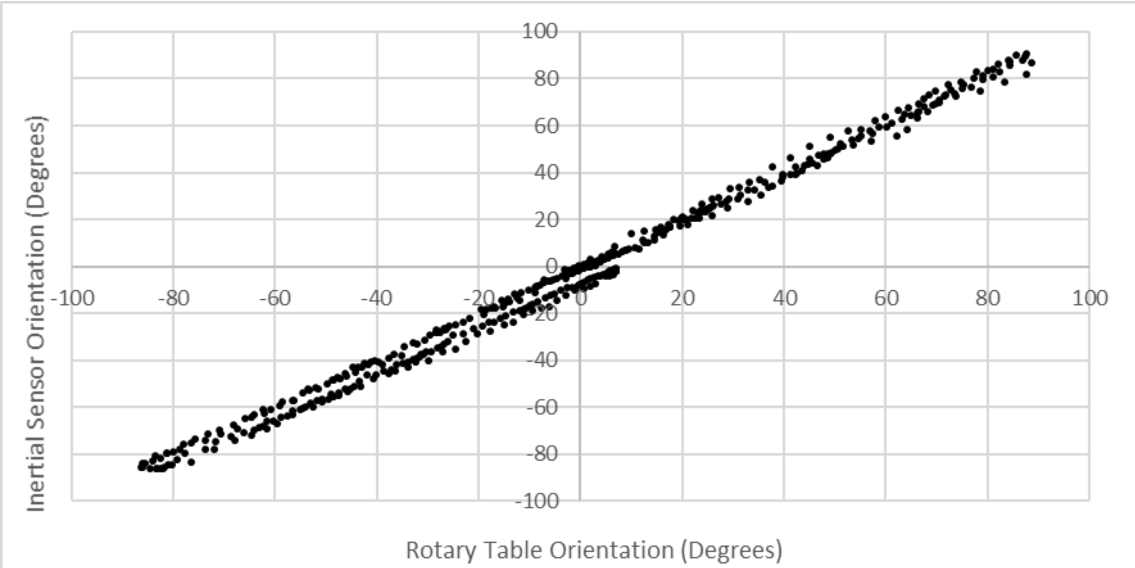


Figure A.8 - Rotary table orientation (x-axis) against inertial sensor 4 orientation (y-axis) for pitch

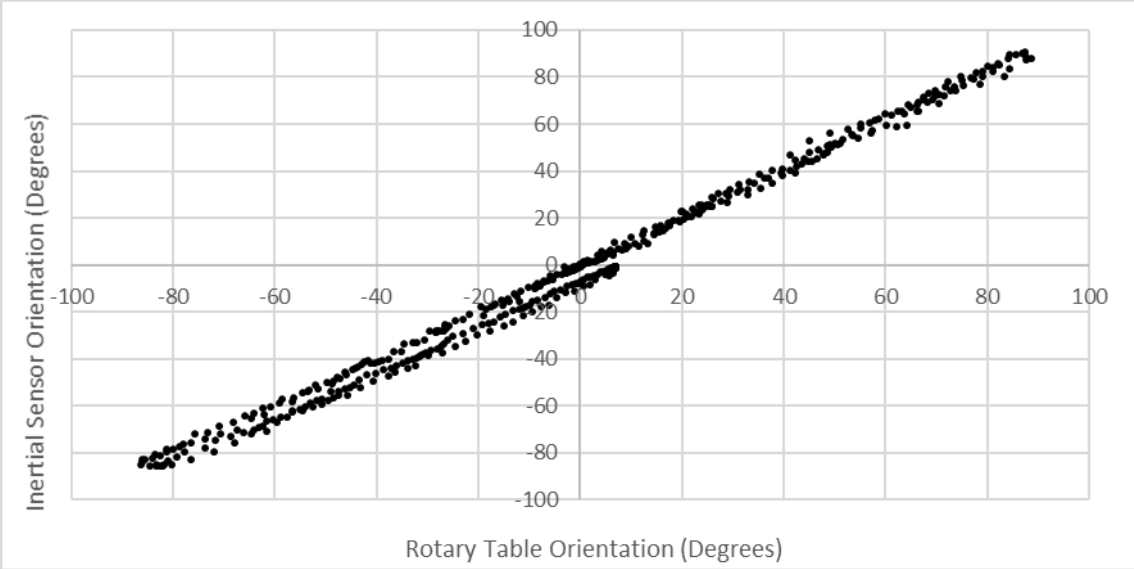


Figure A.9 - Rotary table orientation (x-axis) against inertial sensor 5 orientation (y-axis) for pitch

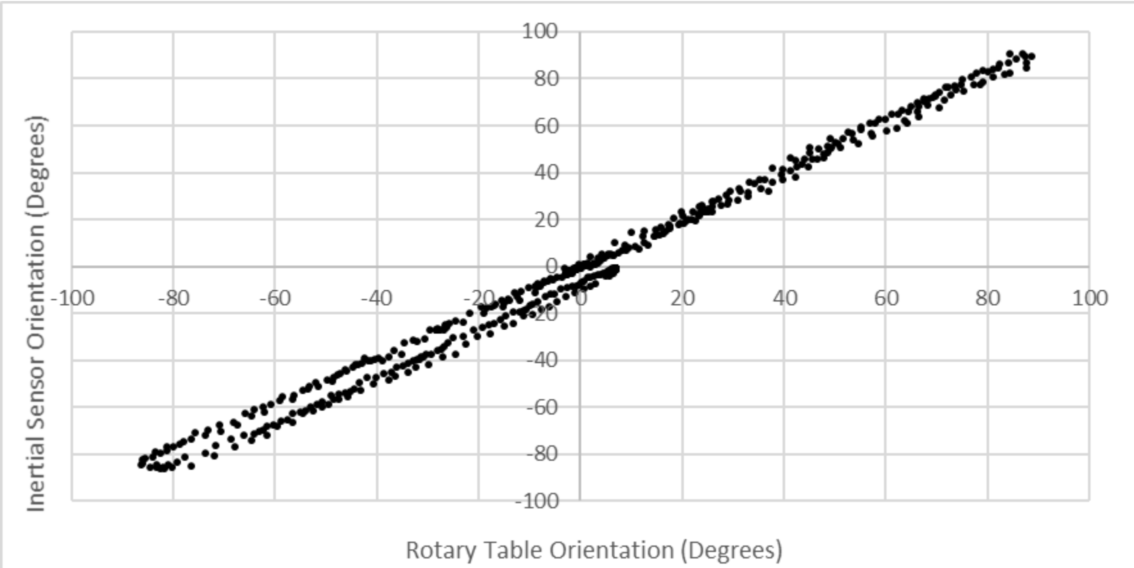


Figure A.10 - Rotary table orientation (x-axis) against inertial sensor 6 orientation (y-axis) for pitch

A.1.1.3. Heading Data

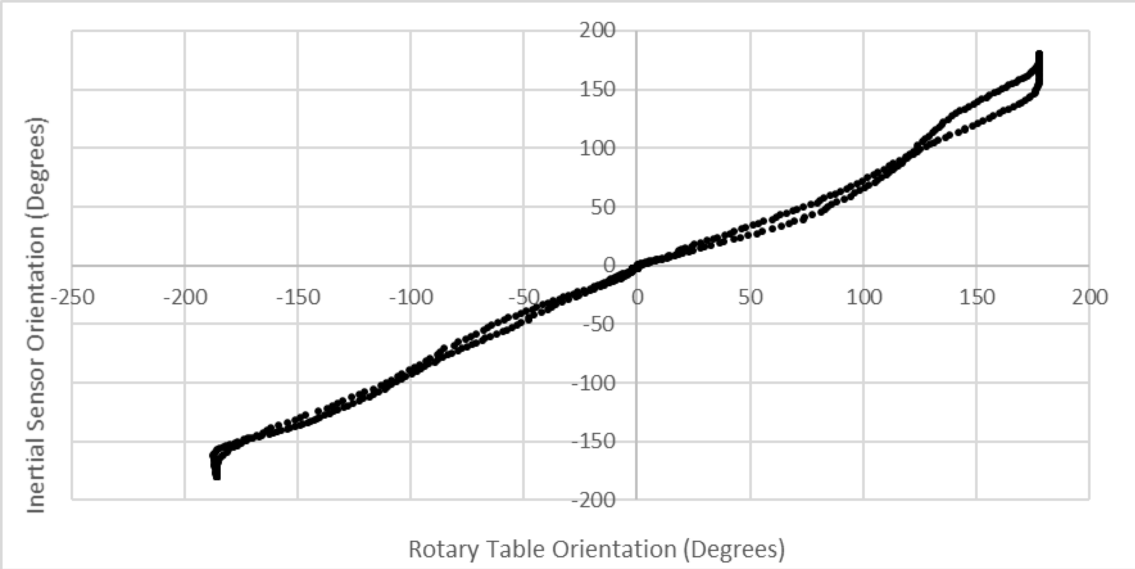


Figure A.11 - Rotary table orientation (x-axis) against inertial sensor 2 orientation (y-axis) for heading

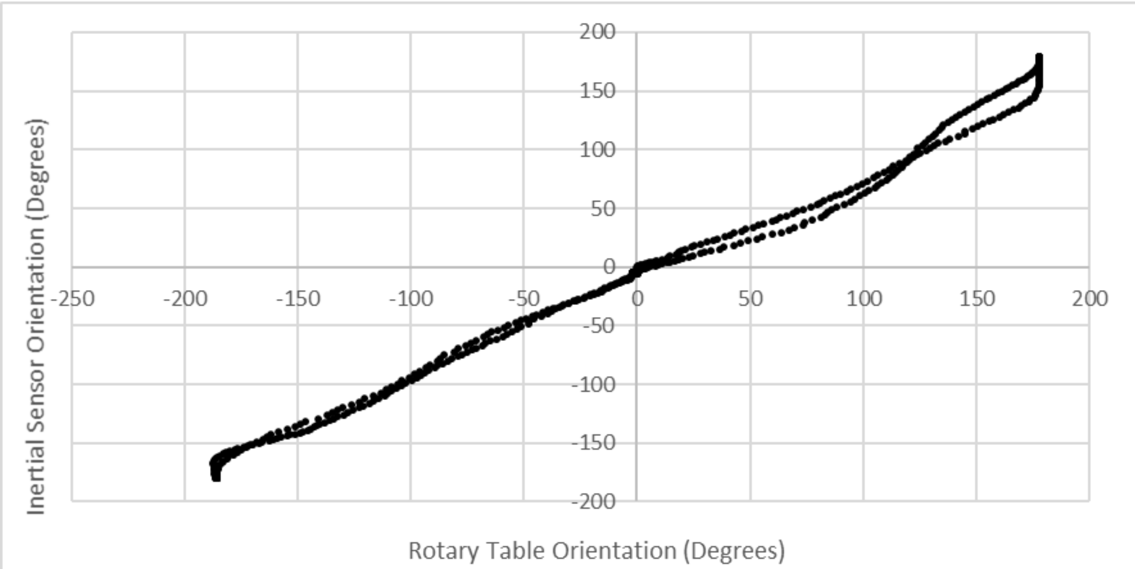


Figure A.12 - Rotary table orientation (x-axis) against inertial sensor 3 orientation (y-axis) for heading

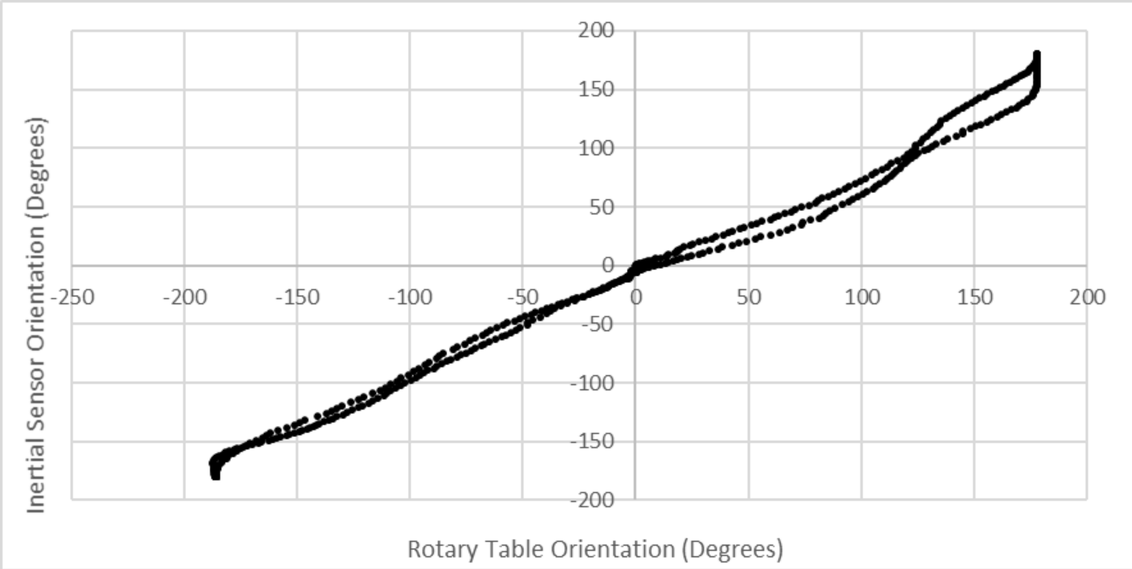


Figure A.13 - Rotary table orientation (x-axis) against inertial sensor 4 orientation (y-axis) for heading

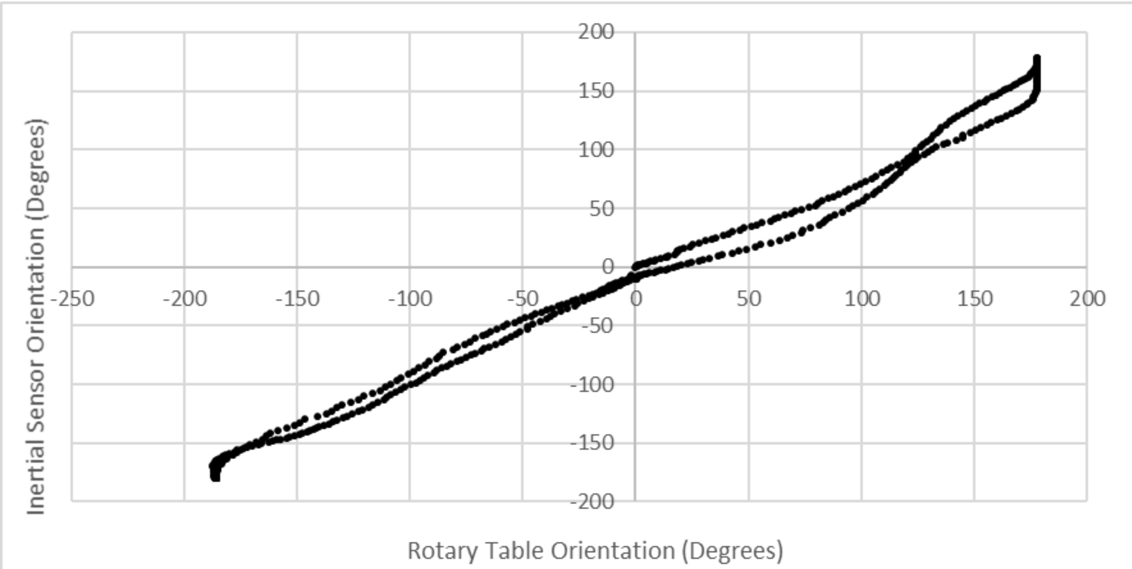


Figure A.14 - Rotary table orientation (x-axis) against inertial sensor 5 orientation (y-axis) for heading

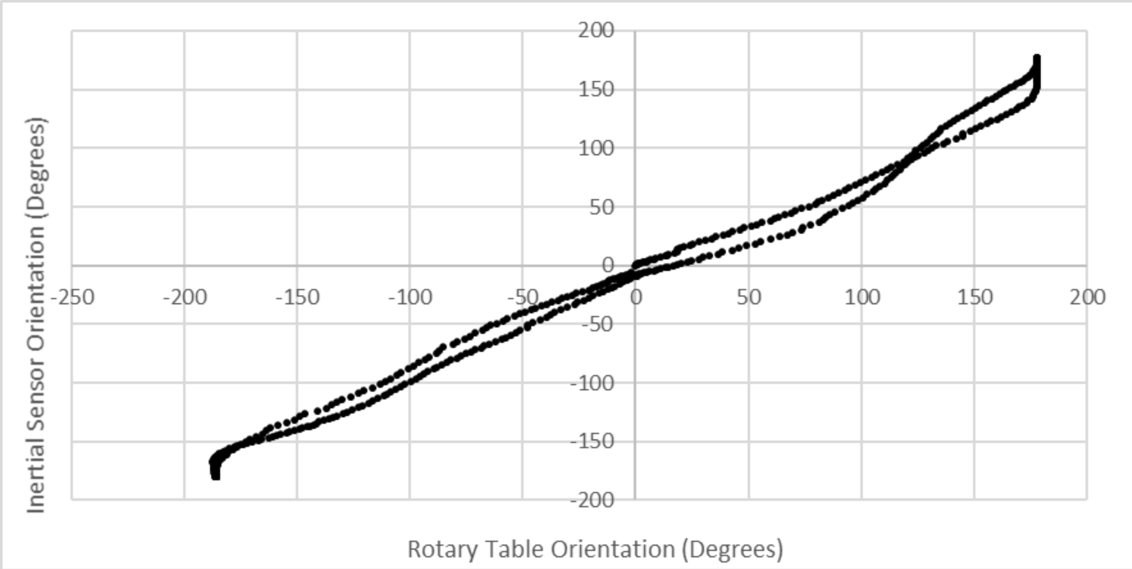


Figure A.15 - Rotary table orientation (x-axis) against inertial sensor 6 orientation (y-axis) for heading

A.1.2. Force-EMG Study Data

A.1.2.1. Correlation Curves

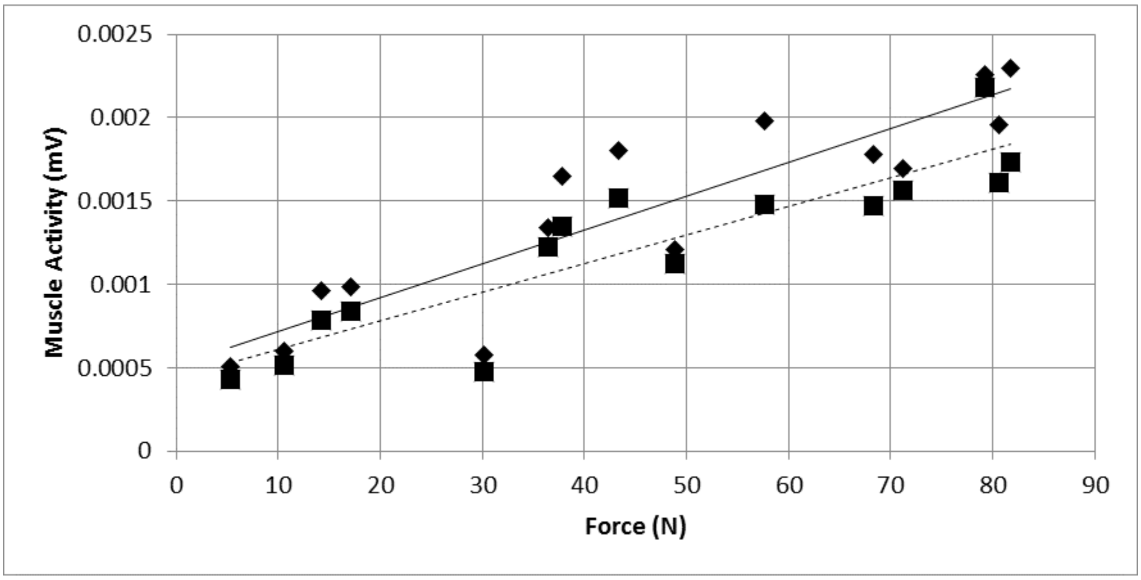


Figure A.16 - Linear relationship of force and EMG for all 3 trials for participant 1. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.802; dotted line – Right side linear trend, R^2=0.802

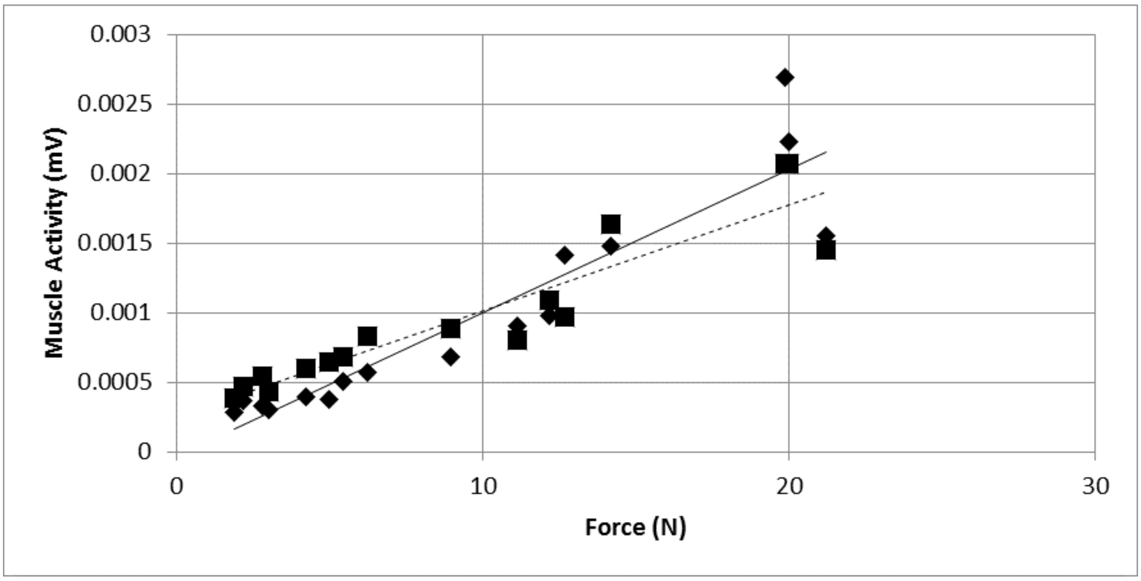


Figure A.17 - Linear relationship of force and EMG for all 3 trials for participant 3. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.867; dotted line – Right side linear trend, R^2=0.862

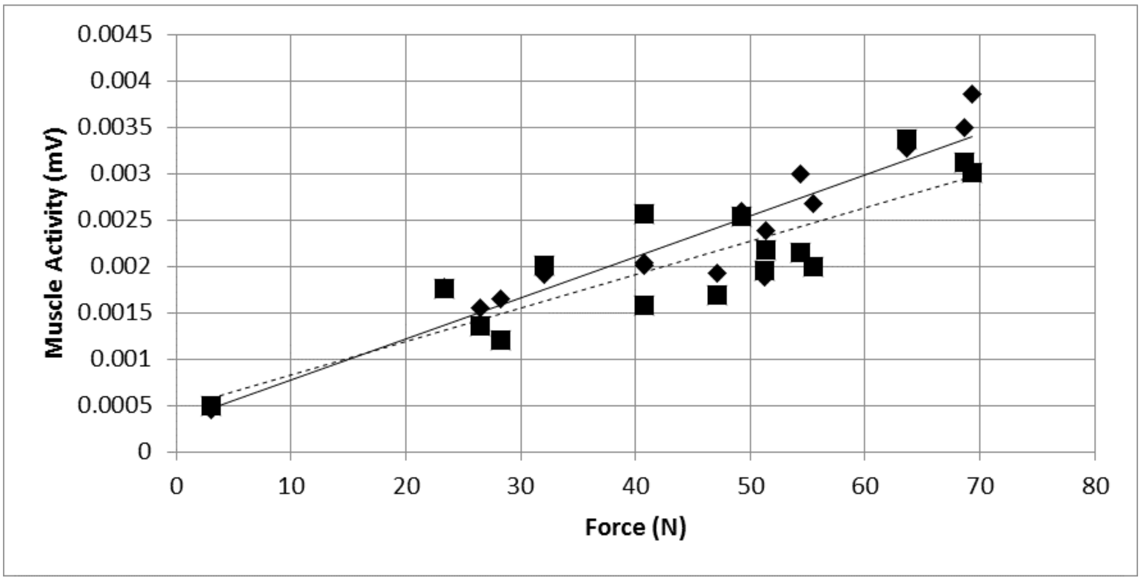


Figure A.18 - Linear relationship of force and EMG for all 3 trials for participant 4. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.875; dotted line – Right side linear trend, R^2=0.846

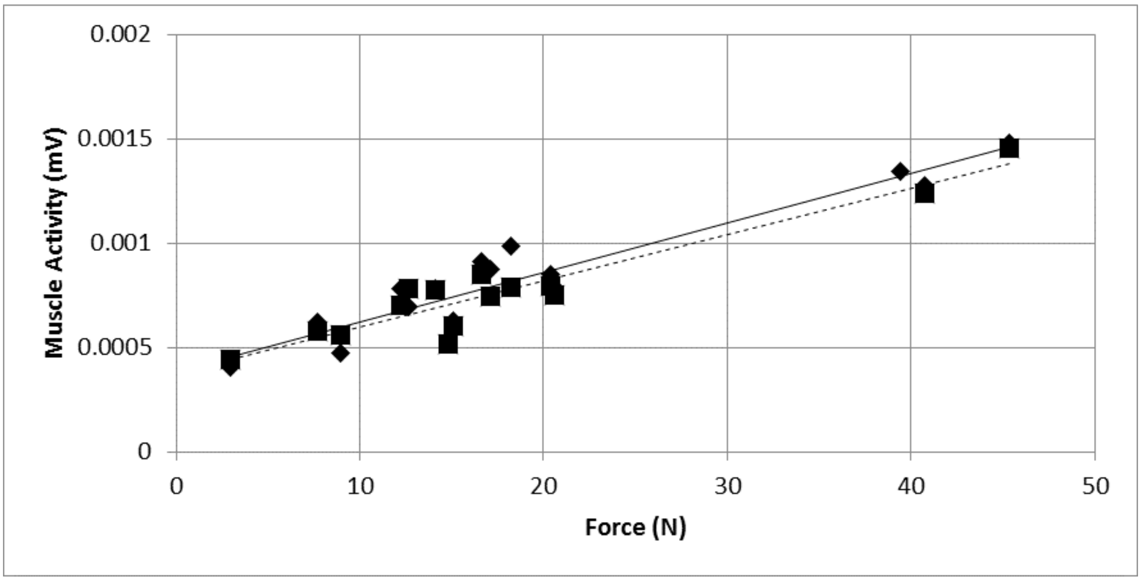


Figure A.19 - Linear relationship of force and EMG for all 3 trials for participant 5. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.889; dotted line – Right side linear trend, R^2=0.897

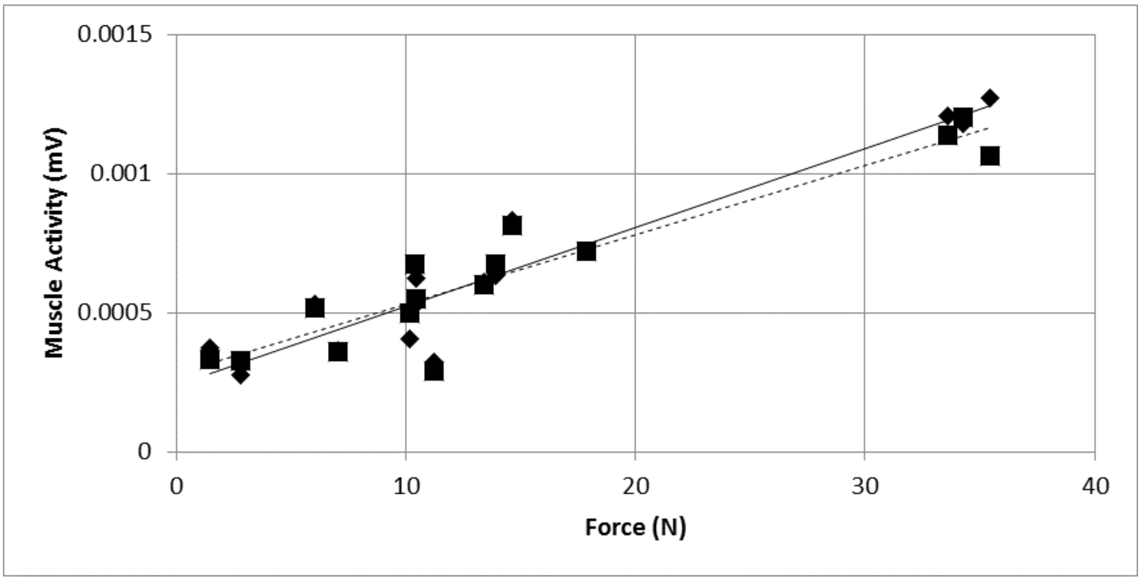


Figure A.20 - Linear relationship of force and EMG for all 3 trials for participant 6. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.907; dotted line – Right side linear trend, R^2=0.873

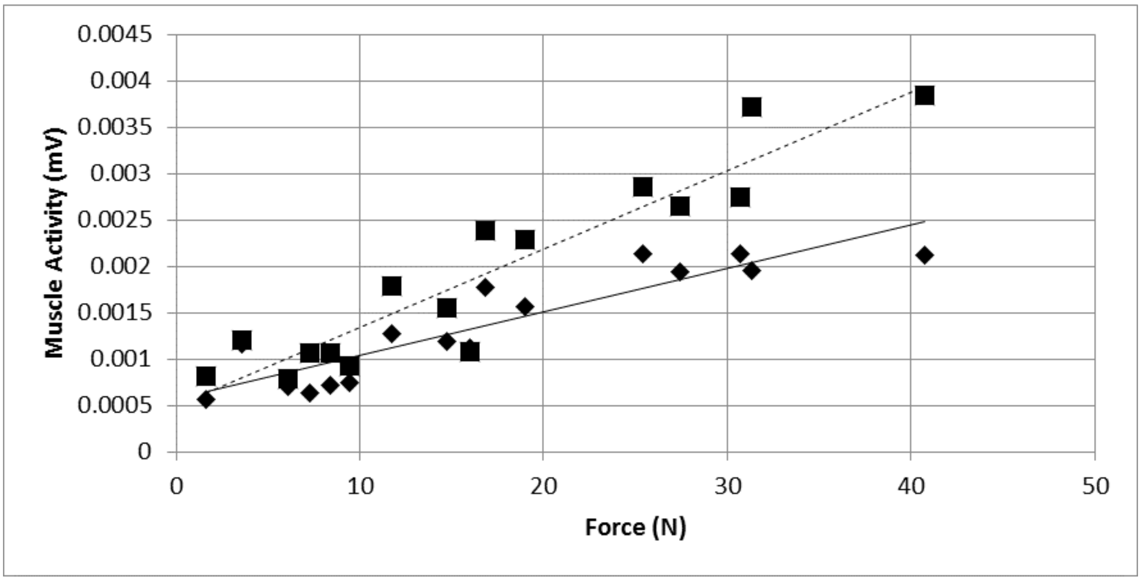


Figure A.21 - Linear relationship of force and EMG for all 3 trials for participant 7. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.816; dotted line – Right side linear trend, R^2=0.878

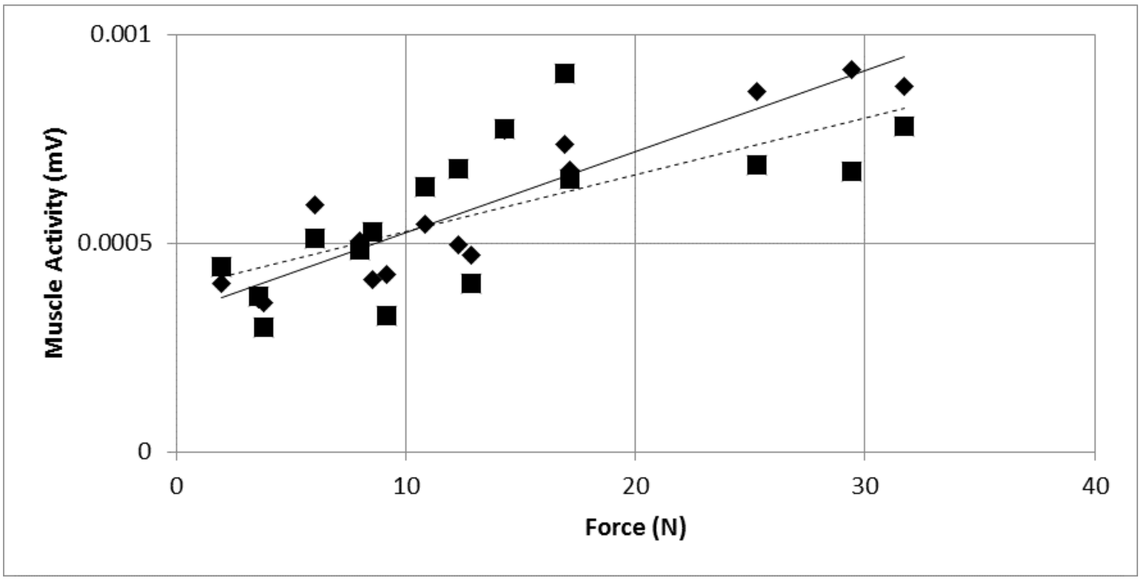


Figure A.22 - Linear relationship of force and EMG for all 3 trials for participant 8. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.826; dotted line – Right side linear trend, R^2=0.472

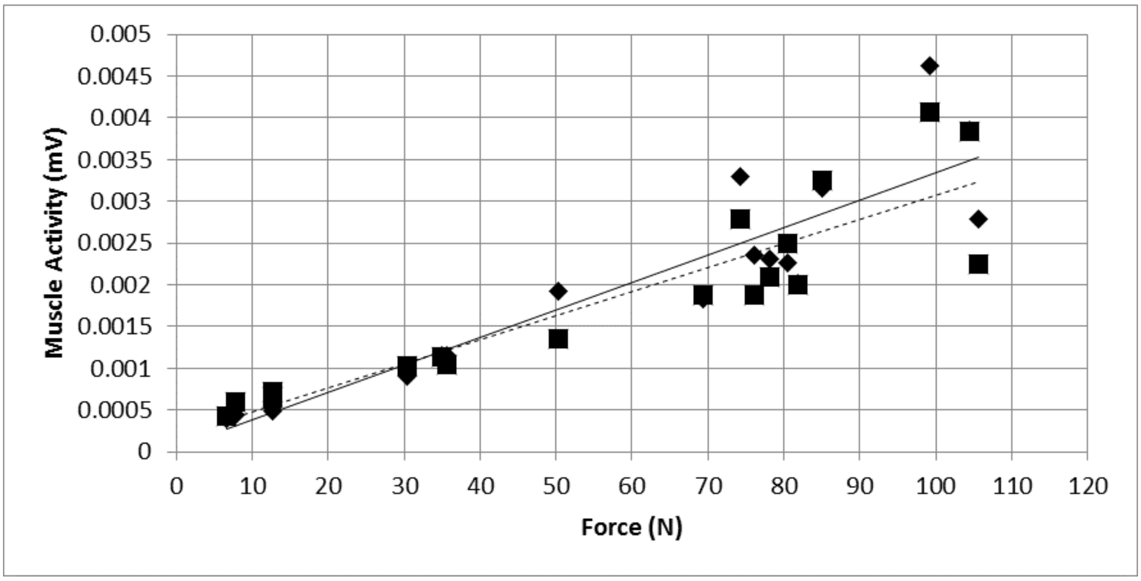


Figure A.23 - Linear relationship of force and EMG for all 3 trials for participant 9. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.833; dotted line – Right side linear trend, R^2=0.808

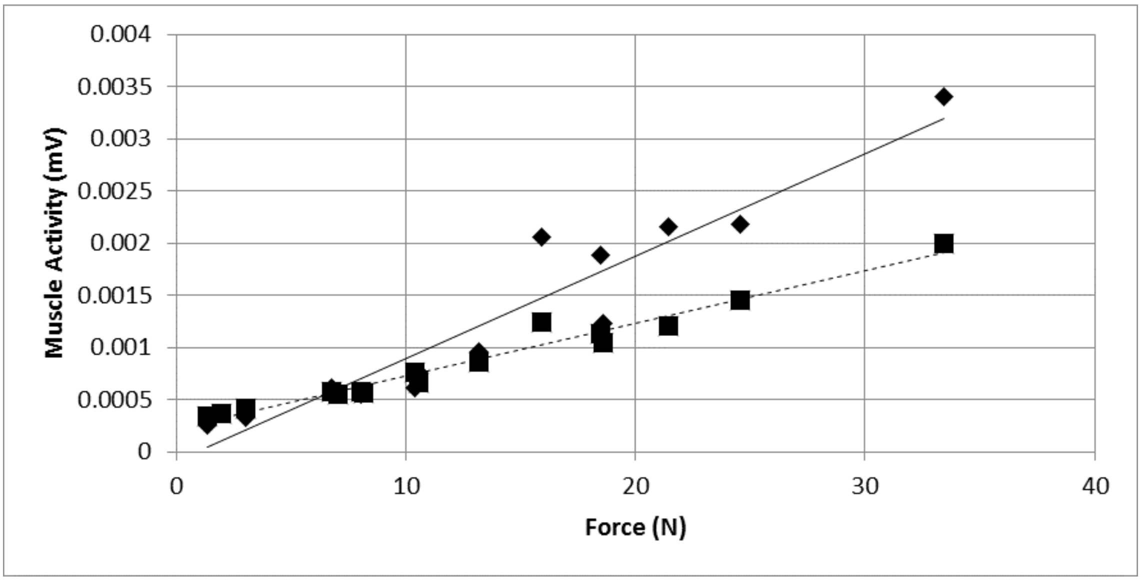


Figure A.24 - Linear relationship of force and EMG for all 3 trials for participant 10. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.913; dotted line – Right side linear trend, R^2=0.967

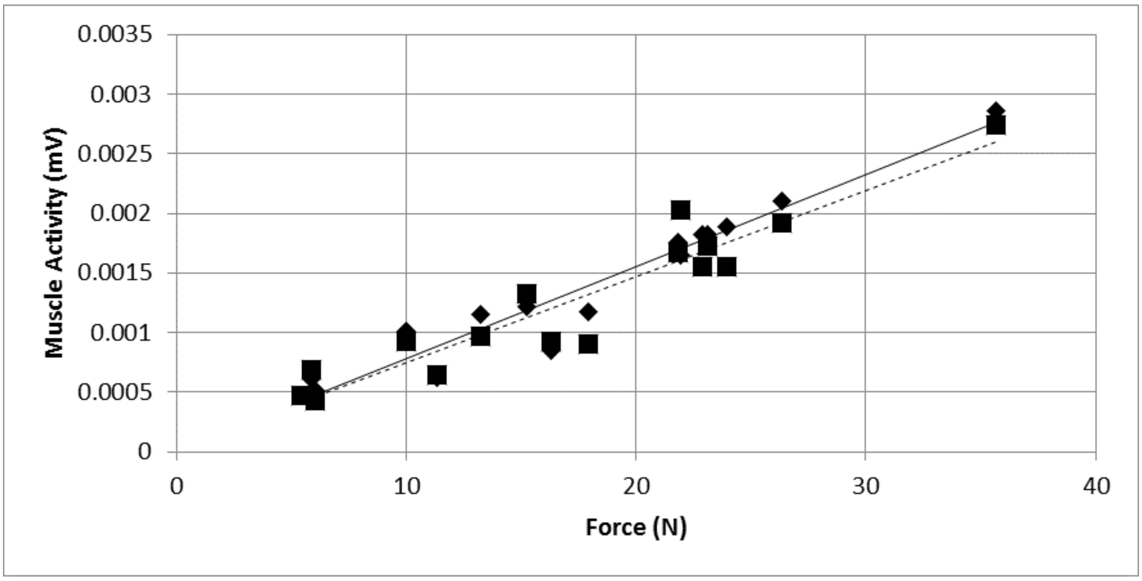


Figure A.25 - Linear relationship of force and EMG for all 3 trials for participant 11. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.940; dotted line – Right side linear trend, R^2=0.892

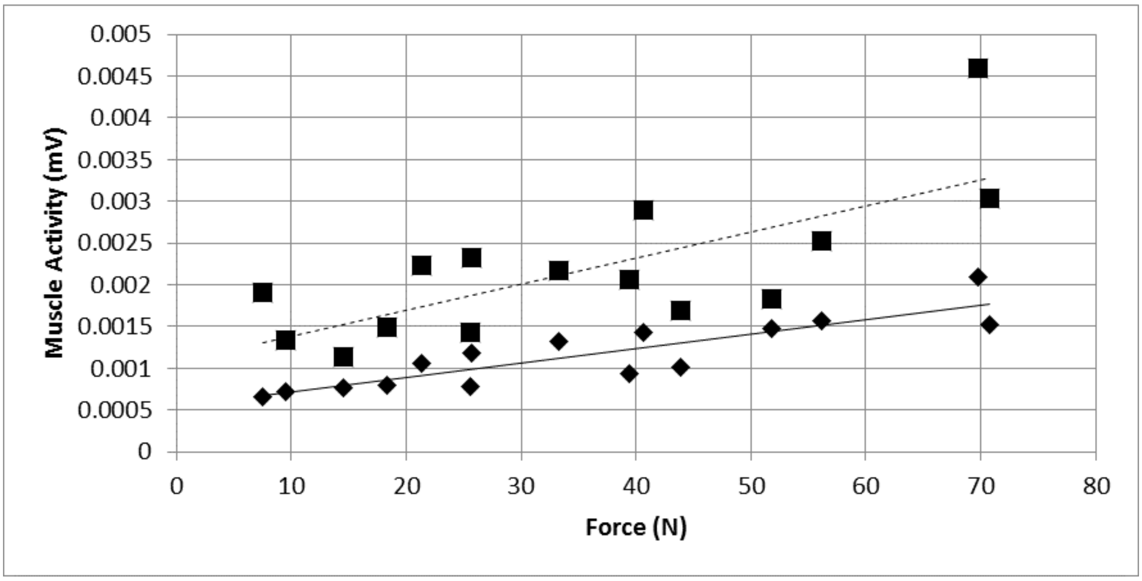


Figure A.26 - Linear relationship of force and EMG for all 3 trials for participant 12. Diamond – Left side; Square – Right side. Solid line – Left side trend line, R^2=0.772; dotted line – Right side linear trend, R^2=0.540

A.1.3. Preliminary Study Data

Table A.1 - Participant Information for Individual Participants

Participant	Age (Years)	Height (m)	Weight (kg)	BMI (kgm^{-2})	Playing Experience (years)	Dominant Side
1	22	1.78	94.0	29.7	15	R
2	21	1.85	111.0	32.4	3	R
3	21	1.78	102.0	32.2	14	R
4	19	1.79	98.0	30.6	5.5	R
5	19	1.83	110.0	32.8	9	R
6	21	1.80	86.0	26.5	11	R
7	22	1.85	96.0	28.0	7	L
8	21	1.82	104.0	31.4	5	R
9	21	1.85	111.0	32.4	3	R

A.1.3.1. ROM Data

Table A.2 – Cervical Segment Peak Active ROM for all 9 Participants

Participant	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
1	62.9	39.0	30.3	19.0	45.5	55.2
2	30.5	42.2	30.9	44.2	12.8	55.7
3	66.2	42.0	39.2	37.2	29.4	42.2
4	45.5	59.5	30.4	24.2	38.9	58.6
5	74.0	10.9	49.1	15.1	14.8	24.1
6	45.2	39.6	40.6	35.2	28.0	29.8
7	43.7	45.5	14.1	22.5	11.6	27.2
8	23.8	15.6	18.9	21.5	8.8	32.5
9	24.2	52.2	6.8	22.0	11.2	11.7

Table A.3 – Upper Thoracic Segment Peak Active ROM for all 9 Participants

Participant	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
1	26.6	20.7	29.6	20.0	27.7	23.5
2	19.4	29.3	9.9	26.8	10.2	26.8
3	25.8	16.3	24.8	20.8	15.3	22.8
4	36.2	23.4	17.0	34.8	18.9	41.0
5	15.8	33.9	13.9	24.3	29.4	45.0
6	48.3	20.4	35.1	23.2	47.5	33.5
7	15.3	38.8	19.7	29.3	57.9	18.2
8	54.7	32.2	22.9	28.5	55.4	35.6
9	36.4	27.1	35.2	32.8	53.8	19.5

Table A.4 – Mid-Thoracic Segment Peak Active ROM for all 9 Participants

Participant	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
1	13.3	33.2	13.1	13.5	16.9	19.6
2	20.0	26.8	14.4	30.4	14.6	31.2
3	25.1	14.6	33.0	18.9	20.0	15.7
4	14.9	17.3	32.8	13.2	21.7	18.6
5	22.3	16.5	20.3	20.2	13.1	23.0
6	49.7	14.8	17.5	32.2	14.3	32.2
7	14.9	50.5	23.5	24.5	18.4	30.3
8	17.0	40.1	20.4	15.9	28.4	11.1
9	39.7	34.0	30.7	54.3	38.5	14.9

Table A.5 – Lower Thoracic Segment Peak Active ROM for all 9 Participants

Participant	Flexion (°)	Extension (°)	Right Lateral Bending	Left Lateral Bending	Right Rotation	Left Rotation

			(°)	(°)	(°)	(°)
1	12.4	54.8	21.3	25.4	43.6	25.4
2	18.6	53.2	43.1	30.9	25.9	30.9
3	2.8	46.7	21.7	35.8	41.3	16.5
4	30.0	38.1	22.1	27.4	35.1	28.6
5	30.0	28.3	43.1	26.4	43.6	16.5
6	52.2	10.0	55.0	36.7	17.8	48.7
7	11.8	41.1	38.8	22.5	13.7	23.3
8	10.4	26.7	10.4	15.5	19.5	18.5
9	22.5	40.6	31.7	20.1	21.8	52.9

Table A.6 – Lumbar Segment Peak Active ROM for all 9 Participants

Participant	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
1	52.4	6.5	30.0	17.8	29.2	37.2
2	38.8	8.4	18.1	20.4	31.3	20.8
3	59.4	9.2	38.6	14.2	34.9	21.9
4	56.0	9.6	23.7	63.1	22.2	57.8
5	68.4	6.0	40.4	28.1	19.2	24.7
6	59.5	7.2	20.5	56.4	81.9	33.7
7	69.6	4.7	49.1	55.0	13.9	35.3
8	26.3	7.5	12.4	20.1	19.6	26.5
9	41.5	8.9	27.6	41.4	44.1	27.9

A.1.3.2. Machine and Live Scrummaging Peak ROM Data

Table A.7 – Peak Cervical ROM for all Participants

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	21.1	40.4	30.5	20.5	18.2	17.3	12.6	10.0	17.1	32.3	13.2	27.0
2	20.8	9.8	36.3	35.2	29.1	18.7	5.9	8.4	10.1	9.0	40.2	45.3
3	27.7	37.8	35.7	30.5	29.7	26.2	7.9	6.5	13.5	18.2	28.4	34.9
4	41.7	32.9	56.2	39.6	9.0	18.5	29.4	25.0	10.1	21.6	43.5	50.4
5	18.8	52.8	40.5	6.2	20.6	27.3	21.2	24.0	24.1	17.0	32.5	23.7
6	20.5	32.9	38.5	28.3	11.9	23.8	12.5	32.2	23.7	18.9	17.3	25.9
7	25.0	43.7	12.9	45.5	27.5	14.1	29.5	22.5	22.6	11.6	29.5	27.2
8	23.2	27.8	12.8	15.6	26.4	18.9	16.7	21.5	41.5	8.8	9.8	32.5
9	N/A	24.2	N/A	52.2	N/A	6.8	N/A	22.0	N/A	11.2	N/A	11.7

Table A.8 – Peak Upper Thoracic ROM for all Participants

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	11.2	9.9	13.4	15.4	4.5	18.4	20.0	8.9	14.5	12.8	11.9	6.4
2	18.1	4.8	36.9	18.2	19.7	7.3	23.7	6.8	7.7	6.6	38.2	16.6
3	16.4	10.7	9.1	15.1	26.4	12.5	6.5	7.3	0.6	8.4	18.7	14.4
4	9.7	10.9	16.7	15.5	10.9	6.0	7.1	23.5	15.5	11.4	12.8	31.0
5	18.7	13.2	19.7	30.9	23.2	9.3	11.8	14.4	14.5	6.1	15.3	48.1
6	17.7	38.3	7.7	2.3	15.6	25.4	5.7	17.2	19.7	21.8	15.9	14.4
7	43.7	10.2	46.2	38.8	38.6	12.2	19.6	29.3	7.8	31.2	57.2	8.2
8	26.9	23.9	24.2	12.2	14.8	19.4	14.4	11.8	9.5	20.6	23.7	30.6
9	27.7	21.3	31.0	15.3	21.3	17.2	24.1	25.1	22.0	36.8	18.1	8.0

Table A.9 – Peak Mid-Thoracic ROM for all Participants

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	1.7	1.5	27.0	22.4	9.5	8.9	18.3	10.9	12.6	12.9	14.8	11.5
2	3.4	0.0	32.9	23.0	25.0	11.9	17.5	5.7	1.7	1.8	21.4	23.1
3	4.0	13.5	11.7	8.4	22.2	14.8	13.7	8.1	16.2	12.1	25.2	7.3
4	27.6	11.1	1.8	13.9	17.1	26.5	7.2	6.4	11.0	14.2	8.7	14.8
5	26.4	19.2	21.0	11.8	11.7	19.1	13.3	12.2	15.1	3.1	22.2	21.2
6	24.7	29.7	24.6	6.7	2.1	5.6	29.4	24.6	21.2	16.4	24.7	19.2
7	0.9	3.7	31.3	47.4	12.3	10.3	11.4	21.4	11.2	6.8	10.8	23.9
8	4.9	1.4	25.8	26.0	6.6	11.8	9.1	7.1	10.7	11.9	17.3	14.8
9	21.9	17.4	4.0	43.6	24.6	9.6	4.8	22.3	15.1	14.5	19.0	8.8

Table A.10 – Peak Lower Thoracic ROM for all Participants

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	8.2	0.0	33.5	44.6	18.0	11.3	6.4	13.4	12.7	26.9	26.2	8.0
2	0.0	0.0	29.1	40.0	9.6	29.3	20.4	3.6	13.9	15.1	19.4	9.3
3	1.9	2.2	39.6	30.1	5.5	7.8	13.7	15.7	5.8	16.3	14.9	11.2
4	32.5	19.8	3.6	26.8	10.0	8.6	17.4	15.6	7.4	14.8	15.2	21.9
5	25.0	9.8	5.1	19.4	14.8	17.8	14.9	20.3	16.4	28.9	14.4	8.0
6	29.3	33.0	11.4	2.5	24.3	19.4	8.2	16.0	8.8	6.5	26.6	30.9
7	1.3	0.6	35.2	35.5	18.7	18.8	5.9	10.6	8.4	9.1	14.3	19.3
8	2.8	2.1	20.8	17.9	4.7	5.4	5.9	10.5	7.4	10.3	7.4	8.0
9	27.8	22.1	0.0	21.3	7.0	13.0	14.8	14.2	16.9	11.6	3.9	28.0

Table A.11 – Peak Lumbar ROM for all Participants

	Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	35.1	38.8	5.2	1.6	8.7	17.8	7.9	10.8	11.6	16.4	11.9	17.8
2	53.5	36.9	6.9	0.0	16.0	10.0	24.2	8.7	34.4	26.5	7.1	7.3
3	59.1	37.4	0.0	11.2	9.5	10.3	9.7	6.8	27.1	6.4	13.5	9.2
4	42.7	40.9	4.2	3.6	28.7	14.3	9.0	18.1	14.4	13.6	9.2	15.1
5	24.7	32.3	2.2	2.1	4.9	24.5	21.5	14.9	19.2	10.4	29.6	15.2
6	37.7	37.6	0.0	4.9	11.7	14.0	11.2	19.0	5.2	21.2	26.3	13.9
7	35.7	50.5	0.6	3.7	28.2	21.0	5.1	18.9	11.5	6.3	9.2	22.0
8	19.1	27.4	6.3	3.2	8.0	23.6	4.2	12.3	6.9	13.1	8.5	13.4
9	24.8	29.1	0.0	2.5	13.7	2.5	15.6	23.4	8.8	25.8	16.8	2.3

A.1.3.3. Machine and Live Scrummaging Peak Angular Velocity Data

Table A.12 – Peak Cervical Angular Velocity for all Participants

	Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	6.8	25.1	-8.3	-11.5	4.3	6.9	-4.9	-4.5	7.5	17.7	-5.2	-10.8
2	12.3	18.2	-13.3	-7.0	7.7	3.9	-7.3	-2.8	14.5	21.4	-13.0	-7.1
3	20.5	20.3	-21.0	-18.3	9.6	7.3	-7.5	-9.6	7.9	13.6	-8.7	-11.3
4	17.1	14.6	-23.8	-17.8	8.4	11.3	-8.1	-7.7	16.3	12.6	-11.2	-13.6
5	10.2	18.1	-12.1	-24.0	7.1	16.9	-7.2	-28.8	12.6	17.6	-11.3	-11.5
6	10.8	16.0	-11.3	-19.9	6.3	11.0	-5.9	-15.3	14.1	11.7	-7.9	-11.6
7	15.4	18.9	-13.5	-20.6	7.4	17.3	-5.4	-16.5	12.6	17.2	-12.2	-13.5
8	14.6	17.5	-12.3	-19.2	6.4	15.9	-6.2	-14.4	12.4	16.8	-11.5	-14.3
9	N/A	18.2	N/A	-20.7	N/A	17.5	N/A	-19.7	N/A	18.2	N/A	-12.5

Table A.13 – Peak Upper Thoracic Angular Velocity for all Participants

	Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	5.2	4.6	-5.6	-6.4	5.5	4.3	-5.7	-9.4	6.3	3.8	-5.6	-4.4
2	10.8	7.0	-12.6	-3.6	6.2	2.8	-6.9	-2.0	19.5	4.4	-21.6	-3.8
3	23.1	4.8	-15.0	-2.9	12.1	6.1	-10.6	-3.8	18.8	4.9	-13.2	-3.9
4	5.2	7.2	-3.4	-13.3	3.2	11.3	-3.7	-6.3	5.2	6.1	-5.4	-13.6
5	12.3	13.2	-12.4	-10.1	12.7	12.0	-7.7	-10.5	17.0	10.1	-14.8	-11.6
6	5.5	12.8	-4.0	-14.1	5.6	6.8	-6.7	-11.0	8.0	7.2	-8.0	-11.7
7	22.1	26.2	-21.8	-15.2	15.3	19.3	-14.5	-17.7	9.3	26.2	-14.9	-17.0
8	14.3	12.3	-16.0	-10.9	8.7	6.8	-9.4	-10.1	8.2	13.6	-10.7	-12.5
9	11.7	10.8	-8.6	-13.6	11.7	17.3	-11.7	-10.3	14.2	13.8	-10.9	-14.9

Table A.14 – Peak Mid-Thoracic Angular Velocity for all Participants

	Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	8.7	5.6	-8.0	-5.1	6.2	5.3	-7.9	-5.1	10.5	6.2	-8.3	-4.6
2	12.6	2.9	-10.9	-3.8	9.9	3.3	-9.2	-4.5	20.1	4.9	-18.4	-2.2
3	9.9	5.7	-8.6	-6.0	5.8	7.0	-7.3	-5.0	14.5	2.7	-14.2	-2.2
4	4.6	14.7	-1.2	-10.2	2.8	9.7	-4.2	-12.2	3.9	8.8	-3.1	-11.6
5	9.2	11.9	-8.7	-7.3	4.8	11.8	-5.9	-12.2	9.0	5.7	-8.9	-3.1
6	10.7	17.3	-10.8	-11.6	9.4	9.4	-10.1	-12.9	8.0	9.4	-9.6	-19.7
7	9.0	11.2	-8.5	-11.2	5.9	6.9	-5.6	-11.9	8.4	11.0	-6.2	-11.9
8	7.4	5.3	-8.4	-7.4	5.9	6.0	-5.3	-6.8	7.8	7.6	-9.5	-9.4
9	6.9	15.0	-6.7	-18.4	6.2	12.2	-4.6	-11.5	12.2	6.5	-15.6	-5.5

Table A.15 – Peak Lower Thoracic Angular Velocity for all Participants

	Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	11.9	8.7	-12.2	-8.3	6.5	7.2	-5.7	-4.9	8.5	5.5	-10.6	-14.7
2	9.5	5.2	-10.8	-3.1	8.4	3.1	-8.4	-6.8	11.9	2.3	-13.6	-3.1
3	6.7	9.0	-8.4	-6.0	6.1	5.9	-5.0	-7.4	4.1	7.8	-5.1	-6.0
4	8.4	10.9	-8.2	-10.9	4.7	6.0	-4.7	-6.9	7.6	12.1	-4.9	-8.3
5	6.1	21.5	-3.4	-20.0	6.7	8.0	-5.0	-24.7	6.3	10.2	-5.0	-10.0
6	10.6	10.3	-11.7	-8.0	8.8	9.4	-9.1	-7.6	9.3	9.8	-8.6	-7.2
7	6.5	9.2	-5.5	-9.1	6.2	5.1	-6.8	-12.0	6.3	5.9	-5.6	-7.7
8	5.8	4.5	-6.3	-4.7	2.5	4.7	-3.5	-4.4	4.2	5.8	-4.4	-5.1
9	4.9	19.2	-4.4	-6.5	3.7	7.4	-5.4	-5.1	6.3	3.4	-3.5	-8.0

Table A.16 – Peak Lumbar Angular Velocity for all Participants

	Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Participant	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**	**Machine**	**Live**
1	7.4	7.9	-6.4	-10.1	3.7	8.7	-4.6	-7.5	5.6	8.8	-5.4	-6.3
2	10.8	2.7	-12.8	-2.4	10.2	3.3	-8.4	-6.9	11.7	2.7	-10.0	-7.6
3	9.7	7.1	-8.3	-3.5	5.3	3.2	-4.8	-4.7	8.4	3.8	-5.3	-2.5
4	6.9	24.0	-15.2	-21.6	5.6	19.4	-11.5	-10.0	6.1	23.5	-7.0	-22.7
5	4.5	10.7	-4.1	-12.1	3.0	9.3	-1.8	-10.3	8.4	15.1	-2.6	-7.3
6	5.1	7.9	-5.2	-10.1	6.8	8.7	-6.5	-7.5	7.1	8.8	-8.8	-6.3
7	5.1	9.4	-5.3	-9.7	7.0	4.2	-5.9	-6.7	4.4	11.9	-3.6	-3.7
8	4.3	3.3	-5.0	-8.4	2.5	6.8	-2.8	-4.2	2.4	7.7	-4.1	-4.4
9	2.9	6.5	-2.9	-5.6	6.9	14.2	-5.4	-3.4	3.0	2.9	-3.5	-13.3

A.1.4. Playing Surfaces Study Data

A.1.4.1. ROM Data

Table A.17 – Cervical Segment Peak ROM for all Grass and 3G Participants

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
31.1	42.4	37.6	26.7	37.5	30.7	36.3	26.6	34.7	37.8	40.8	25.6
20.9	39.9	62.1	41.3	64.8	36.3	43.8	48.1	34.5	36.5	34.8	27.2
25.4	28.4	27.5	31.6	40.7	42.3	57.3	42.4	44.6	42.2	34.4	30.3
36.9	33.5	33.5	32.6	44.7	37.5	42.9	40.4	32.2	35.4	31.4	31.2
36.2	48.9	35.9	34.6	23.9	42.2	55.5	30.1	53.3	32.8	38.0	35.0
46.4	43.2	26.6	53.9	39.7	47.7	39.6	38.1	36.9	30.5	36.4	37.4
32.9	29.4	39.3	35.7	34.2	54.6	54.0	31.9	43.1	35.4	35.2	34.6
30.3	40.1	23.5	47.7	38.1	33.2	35.9	66.9	35.1	49.1	31.3	44.4
37.5	29.6	34.2	47.0	34.2	47.0	41.0	44.9	33.9	45.0	21.9	38.9
38.3	36.8	21.4	36.0	39.7	38.4	30.1	37.3	45.9	35.3	23.3	31.9
41.2	36.0	33.6	28.5	46.5	27.6	41.3	46.8	47.3	36.7	21.8	35.3

Table A.18 – Upper Thoracic Segment Peak ROM for all Grass and 3G Participants

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G
11.1	15.8	30.9	15.8	17.9	19.3	22.8	18.1	27.4	35.1	21.6	19.0
10.2	13.6	31.7	49.1	11.1	24.3	27.2	19.5	10.7	11.7	14.9	31.7
11.9	13.7	39.5	17.6	17.7	13.5	19.2	42.9	17.1	10.3	42.8	36.2
15.7	14.5	48.8	73.1	20.0	18.9	73.5	24.5	40.6	19.8	24.9	36.3
34.2	12.9	30.8	33.9	33.2	11.1	11.4	22.4	19.7	14.3	47.6	18.3
18.1	12.6	29.4	26.6	10.8	18.0	19.5	26.3	18.1	22.1	25.7	15.0
12.5	36.5	22.2	20.7	34.1	21.8	17.8	24.3	21.5	13.9	59.1	45.5
15.0	16.2	36.5	51.6	26.7	33.6	35.3	18.8	30.8	43.4	10.1	43.1
20.6	19.3	19.1	43.8	17.1	15.1	34.2	31.5	28.5	14.6	16.7	25.8
14.5	17.1	57.9	30.5	22.8	20.3	22.2	18.6	18.2	16.4	16.2	29.9
31.7	22.2	20.9	46.7	25.1	30.0	19.6	20.7	14.7	17.4	11.7	29.1

Table A.19 – Lower Thoracic Segment Peak ROM for all Grass and 3G Participants

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
10.0	23.2	17.8	18.7	17.1	14.3	19.7	15.9	18.2	17.0	19.9	16.2
11.1	19.0	29.4	24.5	10.5	16.3	17.3	14.4	23.5	13.5	22.4	10.7
11.5	17.2	26.1	17.1	13.5	19.4	38.4	15.6	19.8	15.6	23.3	20.8
11.3	21.5	20.4	27.7	12.6	15.3	20.4	18.9	54.0	18.3	30.9	19.8
27.8	21.2	54.3	50.6	11.0	16.8	32.4	17.1	15.1	18.3	21.3	19.0
12.1	20.0	13.4	23.4	14.8	18.0	11.4	13.7	21.8	33.3	20.4	17.0
17.7	29.4	23.6	24.2	16.2	14.4	11.7	33.9	21.7	21.6	42.5	26.3
14.1	56.0	32.3	22.4	29.0	16.0	20.9	44.3	19.8	36.3	15.9	22.5
12.6	13.6	19.2	20.6	19.1	15.8	42.3	20.0	27.3	16.5	23.1	17.3
18.4	10.9	14.9	26.3	15.5	16.6	31.4	11.4	18.9	22.7	23.6	19.1
10.7	9.6	28.4	15.0	13.8	11.9	24.6	22.7	19.7	23.7	22.9	16.2

Table A.20 – Upper Lumbar Segment Peak ROM for all Grass and 3G Participants

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
35.5	42.4	2.6	5.1	13.8	18.3	19.2	15.6	18.0	13.4	12.2	11.6
41.6	26.3	4.7	4.6	16.7	11.4	13.5	11.0	19.0	11.3	13.1	19.8
45.0	31.6	6.1	6.1	11.9	17.1	13.6	12.3	15.1	14.1	18.2	16.5
52.0	36.1	5.8	7.5	15.5	13.8	19.8	12.8	18.6	15.4	20.9	15.5
44.0	43.3	3.2	5.3	19.1	17.9	28.1	17.0	17.6	14.6	29.3	17.1
41.8	43.1	3.7	6.9	12.0	14.6	18.6	19.2	35.2	10.7	20.8	14.4
37.2	53.2	6.6	5.6	26.6	18.5	20.1	20.5	31.9	36.7	19.8	20.0
41.2	31.8	5.6	7.7	15.4	20.2	11.9	14.6	25.3	16.2	12.6	22.1
32.8	79.6	7.2	8.1	11.6	13.2	27.1	17.3	15.1	14.0	17.7	18.0
57.0	63.6	4.9	6.7	19.5	13.6	14.2	12.8	17.6	12.8	18.9	13.5
41.9	72.1	7.0	5.8	14.8	19.5	17.9	17.4	20.4	11.6	13.4	17.7

Table A.21 – Lower Lumbar Segment Peak ROM for all Grass and 3G Participants

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
29.4	20.3	10.0	37.4	10.3	22.0	16.1	20.9	12.0	11.9	13.1	23.1
20.1	17.9	11.8	16.0	11.6	20.7	18.2	16.5	12.3	22.9	13.4	23.7
12.4	24.3	10.2	18.3	12.7	15.5	12.6	12.3	20.5	22.4	36.0	15.0
18.3	21.2	11.4	18.4	15.6	14.9	13.1	11.8	19.4	20.0	19.1	21.8
36.7	8.0	14.9	22.4	22.2	11.4	19.7	17.1	20.1	13.0	34.3	18.0
12.8	20.0	22.7	24.2	21.9	17.4	10.6	16.4	11.2	21.6	19.6	23.2
23.8	46.3	10.0	20.1	26.3	15.4	10.0	32.6	20.5	15.5	23.3	25.2
25.3	45.5	14.1	11.6	20.8	20.9	10.3	19.5	16.4	23.4	26.4	24.1
44.9	11.6	13.8	12.2	32.2	17.3	19.1	18.5	48.4	15.2	23.4	19.4
15.0	11.8	19.2	17.8	20.1	14.7	14.8	19.1	14.4	20.6	21.8	19.2
33.6	12.8	37.2	17.9	13.9	18.0	36.2	16.6	23.9	19.1	52.0	24.8

A.1.4.2. 3G and Grass Surface Peak ROM

Table A.22 – Peak Cervical ROM for all Participants of each Group

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G
22.2	30.3	12.6	4.8	26.8	14.8	18.8	4.7	24.8	5.6	7.7	18.3
13.9	6.6	41.4	27.5	43.2	4.2	2.5	32.0	16.3	11.0	23.2	4.8
16.9	18.9	5.0	7.7	7.1	28.2	38.2	8.0	29.7	28.1	2.9	0.2
4.6	22.3	22.3	2.4	9.8	5.0	28.6	26.9	14.8	23.6	7.6	0.8
24.1	32.6	23.9	9.7	15.9	28.1	37.0	6.7	35.5	15.2	25.3	17.3
30.9	2.8	4.4	35.9	13.1	31.8	6.3	5.4	24.6	7.0	10.9	24.9
8.6	19.6	26.2	23.8	16.1	36.4	36.0	7.9	28.7	23.6	10.1	16.4
20.2	6.7	9.0	31.8	18.7	8.8	10.6	44.6	16.7	32.7	7.5	29.6
25.0	19.7	22.8	31.3	16.1	31.3	27.3	9.9	15.9	30.0	14.6	5.9
25.5	11.2	7.6	24.0	19.8	5.6	13.4	18.2	30.6	23.5	2.2	7.9
14.1	17.3	22.4	12.3	31.0	18.4	27.5	31.2	31.5	11.1	14.5	23.5

Table A.23 – Peak Upper Thoracic ROM for all Participants of each Group

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
0.8	11.3	22.1	11.3	12.8	13.8	16.3	5.8	19.6	25.1	15.4	13.6
6.8	2.4	21.1	32.7	7.4	16.2	18.1	6.3	7.1	1.1	9.9	21.1
7.9	9.1	26.3	11.7	11.8	2.3	12.8	28.6	4.7	0.2	28.5	24.1
3.8	3.0	32.5	48.7	0.0	5.9	49.0	16.3	53.7	6.5	16.6	24.2
22.8	1.9	20.5	22.6	22.1	7.4	7.6	14.9	13.1	9.5	31.7	12.2
5.4	1.7	19.6	4.4	7.2	5.3	6.3	17.5	5.4	14.7	17.1	3.3
8.3	24.3	8.1	13.8	22.7	14.5	5.2	16.2	1.0	2.6	39.4	30.3
3.3	10.8	24.3	34.4	17.8	22.4	23.5	12.5	20.5	28.9	6.7	28.7
13.7	6.2	12.7	29.2	11.4	3.4	22.8	21.0	19.0	9.7	11.1	17.2
3.0	4.7	38.6	20.3	15.2	13.5	8.1	5.7	12.1	10.9	10.8	19.9
21.1	14.8	13.9	31.1	16.7	20.0	6.4	7.1	3.1	11.6	7.8	19.4

Table A.24 – Peak Lower Thoracic ROM for all Participants of each Group

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G	Grass	3G
0.0	0.0	12.7	6.2	12.2	3.1	14.1	4.2	13.0	5.0	7.1	4.4
0.0	6.0	19.6	3.0	7.0	4.2	11.5	2.9	9.0	2.3	1.6	7.1
1.0	4.8	17.4	4.7	2.3	0.9	25.6	10.4	13.2	10.4	15.5	0.5
1.0	14.3	13.6	5.1	1.7	3.5	13.6	5.9	49.3	5.5	20.6	13.2
18.5	0.0	36.2	33.7	7.3	1.2	21.6	11.4	3.4	5.5	14.2	6.0
1.4	0.0	8.9	15.6	3.2	5.3	7.6	9.1	14.5	22.2	13.6	2.0
11.8	19.6	2.4	2.8	4.1	9.6	7.8	22.6	0.0	14.4	28.3	17.5
2.7	37.3	21.5	0.0	19.3	4.0	13.9	29.5	6.5	24.2	10.6	15.0
8.4	2.4	12.8	13.7	12.7	10.5	28.2	13.3	18.2	11.0	15.4	11.5
5.6	0.0	9.9	17.5	10.3	4.4	20.9	7.6	5.9	1.8	15.7	12.7
0.0	6.4	18.9	10.0	2.5	7.9	16.4	15.1	13.1	15.8	1.9	4.1

Table A.25 – Peak Upper Lumbar ROM for all Participants of each Group

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
29.6	35.3	0.0	0.0	11.5	6.9	16.0	4.7	6.7	2.8	10.2	1.3
34.7	21.9	0.0	0.0	5.6	9.5	2.9	0.8	7.5	1.1	2.6	8.2
37.5	26.3	0.0	0.0	1.6	5.9	11.3	1.9	12.6	3.4	6.8	5.4
43.3	30.1	0.0	0.0	4.6	3.2	8.2	2.3	15.5	4.5	17.4	4.6
36.7	36.1	2.7	0.0	15.9	6.6	23.4	5.8	14.7	3.8	24.4	5.9
34.8	35.9	0.0	0.0	1.7	3.8	7.2	7.7	29.3	8.9	0.7	3.7
31.0	44.3	0.0	0.5	22.2	15.4	0.0	17.1	26.6	30.6	0.7	16.7
34.3	26.5	1.3	1.4	12.8	16.8	9.9	12.2	4.4	13.5	10.5	18.4
27.3	66.3	1.8	0.0	9.7	11.0	22.6	6.1	12.6	11.7	6.4	6.7
47.5	53.0	0.0	0.0	7.9	3.0	11.8	10.7	6.3	10.7	7.4	2.9
34.9	60.1	0.0	0.0	12.3	7.9	6.6	6.2	0.0	9.7	11.2	6.4

Table A.26 – Peak Lower Lumbar ROM for all Participants of each Group

Flexion (°)		Extension (°)		Right Lateral Bending (°)		Left Lateral Bending (°)		Right Rotation (°)		Left Rotation (°)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
24.5	0.0	0.0	31.2	8.6	1.7	5.1	17.4	10.0	9.9	2.6	2.6
0.0	14.9	9.8	0.0	1.3	0.6	6.8	5.4	1.9	2.4	2.8	3.1
10.3	3.6	8.5	6.9	10.6	4.6	2.2	1.9	0.4	2.0	30.0	4.2
6.9	0.0	1.2	15.3	4.7	4.1	2.6	1.5	16.2	8.3	7.6	1.5
30.6	6.7	12.4	18.7	18.5	9.5	16.4	5.9	0.0	10.8	28.6	6.7
2.3	0.0	18.9	20.2	1.6	6.2	8.8	5.3	1.0	1.3	16.3	19.3
19.8	38.6	0.0	0.0	21.9	12.8	0.0	27.2	0.0	12.9	19.4	21.0
21.1	37.9	3.4	1.3	9.0	17.4	8.6	7.9	13.7	11.2	5.3	20.1
54.1	9.7	3.2	10.2	26.8	6.1	15.9	7.1	40.3	12.7	19.5	7.8
4.2	1.5	16.0	6.5	8.4	3.9	4.0	7.6	12.0	0.5	9.8	7.7
28.0	1.5	31.0	14.9	11.6	6.7	30.2	5.5	19.9	7.6	43.3	4.0

A.1.4.3. 3G and Grass Surface Peak Angular Velocities

Table A.27 – Peak Cervical Angular Velocity for all Participants of each Group

Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
7.9	8.3	-10.3	-7.7	14.5	4.6	-6.9	-6.3	4.8	6.6	-2.3	-5.5
14.1	8.0	-11.4	-7.6	7.1	6.5	-11.4	-6.7	12.9	3.8	-8.4	-5.5
6.0	5.0	-6.8	-6.0	5.4	11.3	-6.1	-9.0	7.3	5.9	-4.7	-2.0
5.5	5.0	-17.7	-3.3	24.4	5.8	-5.2	-7.9	11.5	6.5	-4.8	-5.0
9.0	9.2	-28.6	-19.6	19.7	11.2	-5.6	-7.4	22.1	6.4	-21.9	-8.4
7.2	6.8	-13.9	-10.8	3.4	8.3	-6.3	-6.9	7.7	6.7	-13.7	-9.1
13.5	19.2	-7.6	-10.6	50.4	4.2	-6.8	-12.5	24.1	7.1	-8.9	-13.5
8.5	14.2	-6.1	-12.7	5.5	17.1	-6.4	-4.7	6.5	11.8	-8.0	-17.0
20.7	8.6	-7.4	-11.5	17.7	7.0	-7.7	-8.0	5.9	8.8	-11.9	-6.4
11.0	4.5	-5.2	-7.2	7.6	4.3	-10.3	-5.1	4.6	7.4	-10.2	-7.0
9.3	4.7	-5.3	-6.7	8.4	9.5	-5.7	-8.6	8.2	6.9	-6.7	-10.2

Table A.28 – Peak Upper Thoracic Angular Velocity for all Participants of each Group

Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
9.3	8.5	-3.7	-5.8	7.3	6.9	-6.2	-5.0	6.9	8.9	-10.7	-10.8
7.6	4.8	-7.2	-5.7	7.3	5.4	-3.8	-4.9	4.2	2.3	-3.7	-7.2
6.0	5.0	-5.1	-4.4	7.6	5.8	-8.7	-8.4	6.0	3.1	-7.7	-4.6
24.0	5.5	-7.9	-9.9	21.8	6.3	-7.6	-6.3	39.6	4.6	-45.9	-6.8
8.7	8.6	-6.9	-3.7	5.4	5.4	-14.5	-5.4	14.5	5.2	-13.6	-5.0
9.5	1.3	-3.9	-1.6	3.3	4.8	-6.6	-8.0	10.9	2.7	-4.6	-5.4
3.1	6.3	-5.3	-13.6	4.6	8.9	-22.1	-4.0	10.5	18.8	-28.3	-9.6
8.1	7.6	-7.0	-13.3	6.5	5.4	-9.0	-17.8	6.9	15.9	-4.2	-10.9
6.4	8.6	-7.9	-4.0	10.7	5.7	-12.7	-4.5	7.0	7.1	-5.2	-7.8
24.1	8.2	-5.7	-7.0	6.6	4.1	-5.9	-4.6	6.2	8.8	-5.1	-8.4
4.3	7.9	-6.4	-6.7	7.1	7.2	-6.4	-4.6	3.5	5.7	-3.3	-7.0

Table A.29 – Peak Lower Thoracic Angular Velocity for all Participants of each Group

Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
2.9	1.6	-3.2	0.0	4.1	2.9	-8.0	-1.5	7.9	2.6	-4.3	-2.8
3.2	1.4	-2.3	-1.7	6.2	1.5	-4.1	-2.0	2.8	3.0	-2.0	-1.6
3.9	1.6	-2.2	-2.3	4.1	1.5	-4.6	-2.1	4.5	2.2	-8.7	-1.7
7.6	2.3	-3.8	-0.9	6.1	2.9	-5.8	-3.5	13.1	3.5	-18.4	-8.4
10.2	6.1	-11.8	-3.3	14.4	3.3	-10.0	-3.3	13.9	2.9	-18.1	-3.0
1.8	1.8	-2.3	-3.3	2.6	3.4	-3.2	-2.6	3.1	5.3	-13.3	-5.4
3.7	5.1	-7.8	-5.5	8.6	8.8	-7.5	-6.3	17.7	7.1	-18.3	-6.6
3.9	2.6	-3.4	-2.2	5.2	1.5	-4.2	-1.8	4.6	15.0	-5.6	-2.3
3.4	4.9	-6.2	-6.4	9.9	9.4	-6.4	-7.2	5.2	4.9	-9.0	-4.4
4.7	2.6	-3.0	-2.0	12.2	2.3	-5.6	-2.5	2.9	2.3	-5.6	-6.2
2.8	3.4	-3.6	-3.0	4.9	3.5	-6.8	-3.6	6.6	3.8	-4.4	-5.6

Table A.30 – Peak Upper Lumbar Angular Velocity for all Participants of each Group

Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
5.1	4.1	-3.7	-2.2	6.8	3.0	-4.2	-2.0	4.2	2.0	-7.4	-1.6
1.9	1.5	-2.2	-1.2	3.3	1.8	-1.6	-2.0	4.0	1.5	-3.8	-4.0
2.1	2.6	-2.2	-1.7	3.2	2.4	-2.7	-1.4	5.6	1.8	-4.7	-3.2
4.7	2.1	-6.7	-1.6	4.6	1.6	-3.9	-1.2	13.5	1.6	-11.7	-2.6
10.9	2.5	-8.9	-2.8	8.6	3.0	-18.7	-2.3	19.8	2.4	-12.1	-1.7
7.6	2.2	-2.0	-1.6	2.3	5.3	-4.8	-2.7	2.4	7.4	-15.8	-3.2
8.4	8.0	-3.9	-6.3	3.0	7.5	-7.4	-7.4	2.9	6.0	-13.1	-16.6
6.1	8.3	-4.4	-5.6	6.1	4.5	-3.9	-5.6	4.4	12.4	-5.7	-12.3
6.1	6.0	-7.4	-3.9	8.8	4.3	-7.0	-2.8	3.8	2.4	-4.4	-2.4
3.5	3.1	-7.3	-2.3	3.2	5.8	-5.8	-6.7	2.4	4.3	-4.0	-3.5
2.7	4.4	-1.6	-4.2	5.0	5.8	-4.8	-3.5	2.7	4.3	-3.0	-4.2

Table A.31 – Peak Lower Lumbar Angular Velocity for all Participants of each Group

Flexion (°/s)		Extension (°/s)		Right Lateral Bending (°/s)		Left Lateral Bending (°/s)		Right Rotation (°/s)		Left Rotation (°/s)	
Grass	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**	**Grass**	**3G**
3.5	3.8	-5.2	-3.2	3.3	3.6	-4.7	-6.9	4.0	4.3	-3.1	-3.6
2.0	1.6	-1.1	-2.6	2.3	1.6	-2.6	-1.8	1.7	1.6	-1.3	-2.5
3.8	1.9	-3.2	-2.3	3.3	1.4	-3.6	-1.7	4.2	1.0	-7.3	-1.4
2.6	2.0	-3.2	-2.0	2.2	1.7	-2.3	-1.2	8.3	2.6	-13.1	-1.1
6.6	3.6	-11.5	-5.1	17.6	2.8	-6.5	-3.8	13.8	2.6	-15.4	-3.6
2.5	2.1	-5.0	-2.1	4.0	3.5	-1.3	-3.1	8.5	4.0	-3.2	-7.2
1.6	6.1	-5.6	-8.2	4.2	11.7	-8.5	-7.4	10.8	15.4	-1.8	-7.1
4.1	14.1	-3.9	-9.3	2.9	5.0	-5.1	-3.8	6.7	13.4	-4.9	-6.7
7.4	6.2	-21.8	-4.9	7.5	2.8	-11.1	-4.4	10.4	4.0	-25.5	-5.1
3.1	2.7	-10.4	-2.9	2.0	4.5	-1.9	-3.5	5.2	2.2	-5.3	-2.2
25.6	3.6	-30.7	-3.8	16.5	4.0	-14.9	-2.9	22.4	2.0	-20.3	-2.2

A.1.5. Engagement Techniques Study Data

A.1.5.1. ROM Data

Table A.32 – Peak Range of Motion for Participant 1

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	29.0	9.5	29.9	21.0	48.7	43.1
Cervical (Standing)	22.6	22.0	27.9	23.8	42.2	36.6
Upper Thoracic	21.7	0.0	9.3	5.1	4.4	4.7
Lower Thoracic	9.4	5.1	6.9	13.2	4.6	5.4
Upper Lumbar	23.0	4.1	10.9	12.1	14.7	15.8
Lower Lumbar	40.6	26.1	6.0	7.1	9.5	5.0

Table A.33 – Peak Range of Motion for Participant 2

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	36.9	17.0	32.6	27.5	46.6	42.4
Cervical (Standing)	33.4	28.0	38.9	35.8	43.4	48.5
Upper Thoracic	16.8	0.0	16.3	16.1	42.1	43.5
Lower Thoracic	10.4	0.0	13.8	10.4	15.2	16.1
Upper Lumbar	25.0	11.8	25.8	32.4	38.4	34.0
Lower Lumbar	32.3	31.4	11.9	13.5	14.3	16.0

Table A.34 – Peak Range of Motion for Participant 3

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	49.1	22.3	28.6	19.4	48.3	56.3
Cervical (Standing)	64.9	23.3	33.3	26.9	40.4	46.2
Upper Thoracic	26.1	1.7	6.1	2.0	29.2	16.8
Lower Thoracic	13.8	0.2	9.4	9.2	7.4	6.3
Upper Lumbar	28.6	7.3	14.9	7.1	8.3	6.4
Lower Lumbar	7.9	11.0	7.7	5.7	3.3	3.2

Table A.35 – Peak Range of Motion for Participant 4

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	25.4	10.6	45.1	34.2	44.6	42.8
Cervical (Standing)	56.2	14.2	41.1	34.0	32.4	41.1
Upper Thoracic	25.7	4.3	11.4	19.2	46.6	28.9
Lower Thoracic	5.2	6.0	14.0	13.6	8.6	7.3
Upper Lumbar	34.0	16.5	9.0	11.1	7.7	14.2
Lower Lumbar	23.6	20.7	5.5	6.5	4.9	4.6

Table A.36 – Peak Range of Motion for Participant 5

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	39.2	25.0	26.1	28.7	43.6	49.0
Cervical (Standing)	35.3	41.8	37.1	30.9	38.9	58.4
Upper Thoracic	42.1	0.0	10.9	12.0	32.4	34.6
Lower Thoracic	28.7	16.0	15.5	14.5	26.4	30.1
Upper Lumbar	25.1	9.9	16.9	12.2	18.9	8.3
Lower Lumbar	47.7	14.9	10.5	10.4	12.2	13.0

Table A.37 – Peak Range of Motion for Participant 6

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	38.3	29.2	29.0	31.6	38.8	43.3
Cervical (Standing)	52.4	23.4	34.0	33.8	41.0	45.8
Upper Thoracic	31.1	2.7	13.7	11.5	36.4	30.9
Lower Thoracic	24.7	10.1	17.9	17.7	27.3	23.6
Upper Lumbar	37.0	5.7	12.0	7.3	9.6	11.1
Lower Lumbar	10.1	14.5	3.4	4.5	10.9	13.2

Table A.38 – Peak Range of Motion for Participant 7

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	36.8	39.6	37.1	41.3	42.1	41.9
Cervical (Standing)	40.0	40.1	33.2	45.3	42.1	46.9
Upper Thoracic	28.3	0.0	11.0	17.1	31.4	43.5
Lower Thoracic	24.5	0.0	19.5	18.3	24.3	20.0
Upper Lumbar	43.5	20.2	9.9	7.8	7.3	7.0
Lower Lumbar	26.8	17.4	12.5	17.3	7.6	14.8

Table A.39 – Peak Range of Motion for Participant 8

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	35.9	41.3	33.1	35.8	43.9	50.7
Cervical (Standing)	40.0	38.4	34.5	35.8	49.4	44.0
Upper Thoracic	11.7	0.0	2.7	5.1	23.8	19.1
Lower Thoracic	6.4	1.3	8.5	6.5	21.3	24.5
Upper Lumbar	21.5	3.6	11.1	7.3	6.2	8.4
Lower Lumbar	33.5	11.1	16.3	7.8	9.0	3.1

Table A.40 – Peak Range of Motion for Participant 9

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	31.2	36.6	38.6	34.5	54.8	58.2
Cervical (Standing)	47.3	43.7	37.2	33.8	54.7	53.0
Upper Thoracic	0.0	30.3	10.8	7.0	18.4	20.4
Lower Thoracic	24.4	14.9	42.1	32.3	42.9	33.6
Upper Lumbar	64.1	9.7	12.4	15.3	21.9	30.1
Lower Lumbar	28.9	24.6	4.4	9.6	9.2	8.8

Table A.41 – Peak Range of Motion for Participant 10

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	35.6	28.3	42.4	30.7	49.8	45.6
Cervical (Standing)	42.5	47.8	43.2	39.3	57.3	39.9
Upper Thoracic	15.0	14.1	11.3	9.4	46.2	35.9
Lower Thoracic	12.5	15.1	14.8	18.8	37.6	32.3
Upper Lumbar	29.1	6.2	17.0	12.6	3.0	9.8
Lower Lumbar	45.7	21.7	5.2	9.6	8.5	9.6

Table A.42 – Peak Range of Motion for Participant 11

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	43.5	18.9	18.4	18.6	50.9	39.4
Cervical (Standing)	49.3	25.5	25.4	29.6	44.6	45.6
Upper Thoracic	10.6	1.1	5.8	5.2	16.7	20.3
Lower Thoracic	11.4	4.3	9.3	11.6	43.9	32.2
Upper Lumbar	30.3	6.3	15.1	11.2	5.4	6.4
Lower Lumbar	31.7	18.8	12.8	12.7	7.9	9.0

Table A.43 – Peak Range of Motion for Participant 12

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	54.0	39.5	28.5	28.5	38.3	44.5
Cervical (Standing)	50.4	46.9	31.7	34.9	36.4	45.3
Upper Thoracic	14.4	4.6	8.7	8.4	17.2	18.1
Lower Thoracic	20.7	0.0	16.3	18.3	35.1	36.7
Upper Lumbar	35.2	7.9	16.8	15.8	17.4	20.1
Lower Lumbar	20.5	17.0	3.0	1.9	8.2	16.4

Table A.44 – Peak Range of Motion for Participant 13

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	31.6	25.3	20.7	18.3	46.9	44.0
Cervical (Standing)	40.0	32.4	40.3	31.6	39.7	39.2
Upper Thoracic	36.3	0.0	6.6	4.2	13.3	16.8
Lower Thoracic	13.0	4.3	12.5	14.1	33.3	31.0
Upper Lumbar	30.4	28.8	8.6	10.0	4.9	6.9
Lower Lumbar	33.6	16.3	4.9	6.6	6.8	7.3

Table A.45 – Peak Range of Motion for Participant 14

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	35.4	27.9	12.0	12.6	14.1	12.9
Cervical (Standing)	33.6	39.8	26.1	27.9	21.9	26.7
Upper Thoracic	12.0	14.0	9.5	4.2	29.3	26.6
Lower Thoracic	24.2	3.2	12.9	10.3	42.3	60.5
Upper Lumbar	34.9	21.9	17.4	23.3	6.5	4.1
Lower Lumbar	28.8	1.7	8.3	9.2	18.0	11.4

Table A.46 – Peak Range of Motion for Participant 15

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	50.5	40.8	18.6	19.8	37.7	29.1
Cervical (Standing)	45.1	45.1	32.0	37.4	38.0	32.3
Upper Thoracic	37.7	3.9	10.8	10.2	8.1	14.7
Lower Thoracic	24.3	16.9	14.6	18.1	34.0	36.2
Upper Lumbar	30.0	29.0	11.2	15.6	9.2	10.8
Lower Lumbar	16.3	10.8	9.2	8.0	12.5	9.7

Table A.47 – Peak Range of Motion for Participant 16

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	46.1	29.7	26.1	38.0	37.9	50.7
Cervical (Standing)	32.1	33.6	35.1	40.7	34.9	34.0
Upper Thoracic	13.5	2.1	16.2	13.1	33.5	25.1
Lower Thoracic	27.8	11.0	15.7	16.0	39.4	33.5
Upper Lumbar	30.8	5.5	17.0	6.8	12.8	12.0
Lower Lumbar	13.2	35.8	14.3	9.1	4.4	7.8

Table A.48 – Peak Range of Motion for Participant 17

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	33.2	22.3	22.5	35.1	43.6	41.2
Cervical (Standing)	35.7	31.8	34.6	22.4	29.1	34.9
Upper Thoracic	21.6	0.0	7.9	8.3	11.1	8.5
Lower Thoracic	14.4	0.0	8.9	11.0	55.5	33.9
Upper Lumbar	29.6	11.2	12.0	13.5	7.8	8.2
Lower Lumbar	11.7	26.6	17.1	8.9	11.1	7.6

Table A.49 – Peak Range of Motion for Participant 18

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	47.4	15.1	16.1	25.4	37.8	44.5
Cervical (Standing)	57.9	30.1	30.8	28.3	40.7	40.3
Upper Thoracic	33.1	0.0	8.1	8.2	10.2	15.2
Lower Thoracic	5.7	8.6	9.0	24.1	23.6	19.5
Upper Lumbar	44.7	16.4	6.4	8.9	2.6	8.0
Lower Lumbar	10.7	21.3	8.0	4.2	4.1	5.0

Table A.50 – Peak Range of Motion for Participant 19

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	32.2	23.0	33.4	34.7	52.2	43.5
Cervical (Standing)	35.3	26.4	40.8	26.5	46.6	53.1
Upper Thoracic	10.5	0.0	12.3	6.0	14.6	16.6
Lower Thoracic	14.6	11.4	14.7	17.4	41.9	42.9
Upper Lumbar	16.3	11.0	9.2	16.4	15.3	12.0
Lower Lumbar	41.5	40.4	15.3	20.0	11.5	15.2

Table A.51 – Peak Range of Motion for Participant 20

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	51.0	16.7	38.9	41.6	47.4	46.4
Cervical (Standing)	52.0	25.3	38.0	42.6	46.2	46.0
Upper Thoracic	36.8	2.0	2.9	6.3	30.4	27.8
Lower Thoracic	31.3	4.7	20.9	15.9	36.5	41.7
Upper Lumbar	44.6	17.2	14.6	24.7	10.0	13.1
Lower Lumbar	10.1	29.5	9.5	8.6	9.3	11.4

Table A.52 – Peak Range of Motion for Participant 21

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	33.7	5.1	22.1	13.2	38.3	53.3
Cervical (Standing)	51.2	29.0	38.9	23.9	40.1	37.8
Upper Thoracic	1.9	27.2	9.0	3.1	10.2	7.1
Lower Thoracic	11.7	2.9	14.6	11.8	16.4	36.2
Upper Lumbar	41.8	20.0	21.1	13.6	13.6	9.6
Lower Lumbar	14.6	9.9	2.1	1.7	5.5	3.9

Table A.53 – Peak Range of Motion for Participant 22

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	32.1	25.0	22.9	30.4	45.8	41.9
Cervical (Standing)	43.0	29.8	33.9	29.1	58.2	38.2
Upper Thoracic	4.9	27.7	15.1	10.8	38.9	13.1
Lower Thoracic	31.8	11.6	20.7	24.8	23.6	33.6
Upper Lumbar	21.6	19.7	13.1	13.7	6.9	10.8
Lower Lumbar	15.5	57.8	5.8	4.7	20.4	9.8

Table A.54 – Peak Range of Motion for Participant 23

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	45.2	25.0	51.7	34.9	58.2	52.2
Cervical (Standing)	46.2	36.1	34.5	19.8	58.4	57.3
Upper Thoracic	25.1	0.0	9.7	7.1	12.4	15.4
Lower Thoracic	9.3	6.9	19.1	18.6	35.5	26.6
Upper Lumbar	38.3	23.1	14.1	16.7	6.7	6.2
Lower Lumbar	20.7	13.0	4.0	5.2	7.6	10.5

Table A.55 – Peak Range of Motion for Participant 24

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	36.1	29.9	31.3	24.4	32.9	36.2
Cervical (Standing)	46.9	38.5	41.3	35.5	33.1	35.5
Upper Thoracic	14.2	1.6	17.4	15.3	34.0	36.0
Lower Thoracic	28.7	9.8	18.7	18.6	15.5	17.0
Upper Lumbar	26.7	5.7	13.6	11.4	8.9	8.8
Lower Lumbar	28.0	17.0	4.6	5.6	20.4	15.0

Table A.56 – Peak Range of Motion for Participant 25

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	43.9	34.8	36.3	29.0	33.7	32.5
Cervical (Standing)	42.4	58.3	33.1	37.4	36.9	36.6
Upper Thoracic	12.5	0.0	12.7	14.3	33.1	38.5
Lower Thoracic	26.4	8.1	15.8	11.8	17.7	8.9
Upper Lumbar	26.7	0.0	14.4	11.2	12.8	6.2
Lower Lumbar	29.4	20.5	3.5	5.2	13.3	20.4

Table A.57 – Peak Range of Motion for Participant 26

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	34.5	34.1	34.7	34.2	43.2	45.2
Cervical (Standing)	47.0	50.0	39.8	34.1	44.7	38.8
Upper Thoracic	7.7	0.0	9.9	9.3	24.4	27.7
Lower Thoracic	10.7	1.2	9.7	10.4	19.3	24.9
Upper Lumbar	20.9	9.2	12.9	12.3	8.9	8.1
Lower Lumbar	41.0	13.5	11.2	10.0	7.2	10.2

Table A.58 – Peak Range of Motion for Participant 27

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	37.5	27.6	26.0	34.3	46.2	39.0
Cervical (Standing)	47.6	45.7	35.1	35.5	48.3	33.6
Upper Thoracic	6.5	2.9	7.7	11.0	26.3	25.1
Lower Thoracic	10.5	9.8	9.0	12.7	21.4	26.5
Upper Lumbar	27.2	8.2	13.0	12.9	9.0	8.3
Lower Lumbar	32.3	20.6	10.1	8.4	9.4	9.3

Table A.59 – Peak Range of Motion for Participant 28

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	40.1	33.6	24.3	26.4	34.1	40.8
Cervical (Standing)	44.7	39.7	34.6	33.9	33.1	35.1
Upper Thoracic	10.4	0.0	11.2	11.8	25.4	25.7
Lower Thoracic	18.1	0.0	11.9	12.2	26.4	25.4
Upper Lumbar	31.1	6.1	16.4	13.0	14.2	9.1
Lower Lumbar	33.5	22.1	8.7	6.2	11.8	14.0

Table A.60 – Peak Range of Motion for Participant 29

	Flexion (°)	Extension (°)	Right Lateral Bending (°)	Left Lateral Bending (°)	Right Rotation (°)	Left Rotation (°)
Cervical (Scrummaging Position)	25.3	20.5	23.0	21.0	28.1	30.2
Cervical (Standing)	23.0	20.6	26.4	23.4	28.7	34.2
Upper Thoracic	20.7	0.0	9.1	11.5	23.4	30.8
Lower Thoracic	9.9	3.2	9.8	9.8	15.7	26.7
Upper Lumbar	28.0	7.7	14.5	11.6	11.3	9.8
Lower Lumbar	41.4	14.3	8.3	6.0	9.4	12.5

A.1.5.2. Machine Scrummaging Data

A.1.5.5. Kinematic Data Variation

Table A.61 – Coefficient of Variation for all Segments and all Conditions for Participant 2

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	N/A	N/A	N/A	3.3	21.3	20.1	12.7	38.1	22.3	9.2	30.2	16.9	4.9	18.5	44.5
Live (CTPE)	48.0	57.4	103.7	53.2	57.3	73.8	48.7	46.7	56.1	48.7	45.3	54.6	31.2	40.4	45.7

Table A.62 – Coefficient of Variation for all Segments and all Conditions for Participant 3

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	6.6	7.6	0.5	14.7	2.6	14.3	30.7	7.0	21.1	13.0	3.4	3.9	13.0	6.7	1.0
Live (CTPE)	23.2	5.6	10.4	20.1	18.5	29.6	18.7	21.1	6.9	26.0	28.2	17.4	26.2	34.7	36.7

Table A.63 – Coefficient of Variation for all Segments and all Conditions for Participant 4

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	31.6	30.2	24.7	25.3	26.0	22.9	22.0	37.4	20.9	49.3	28.4	33.3	44.4	69.4	10.4
Live (CTPE)	23.8	48.7	47.4	28.9	22.7	19.5	39.4	59.9	36.6	42.8	47.8	66.9	33.0	58.7	6.6

Table A.64 – Coefficient of Variation for all Segments and all Conditions for Participant 5

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	4.0	4.1	29.6	13.8	14.9	25.0	15.2	10.5	9.4	12.9	6.9	72.9	24.3	20.8	26.2
Live (CTPE)	7.6	12.6	34.0	23.1	16.8	24.0	22.4	4.3	13.3	11.7	11.8	46.6	11.6	24.6	39.9

Table A.65 – Coefficient of Variation for all Segments and all Conditions for Participant 6

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	43.4	15.1	19.8	28.6	22.1	6.6	10.4	8.4	8.9	20.0	20.4	64.5	30.0	25.8	12.3
Live (CTPE)	43.3	22.8	33.6	28.6	22.2	15.2	28.3	30.5	31.5	11.7	23.1	37.0	14.4	11.7	32.1

Table A.66 – Coefficient of Variation for all Segments and all Conditions for Participant 7

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	22.0	10.4	5.9	19.2	4.3	3.4	28.7	22.7	25.5	13.5	14.4	26.1	12.3	14.8	23.8
Live (CTPE)	19.6	9.1	30.0	17.7	18.7	13.8	18.5	30.1	21.6	30.8	29.5	54.5	23.9	12.4	51.4

Table A.67 – Coefficient of Variation for all Segments and all Conditions for Participant 11

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	37.9	35.6	8.9	39.4	33.2	32.8	23.8	33.7	34.6	14.1	45.1	38.2	28.5	61.0	25.8
Live (CBS)	29.0	27.0	24.2	8.4	3.3	15.1	24.1	8.6	16.8	9.7	24.0	20.0	14.5	14.9	36.1
Live (CTPE)	2.8	21.2	3.0	16.6	31.2	2.1	20.1	49.5	19.7	83.4	60.0	75.5	81.4	67.9	90.9

Table A.68 – Coefficient of Variation for all Segments and all Conditions for Participant 12

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	32.9	51.7	51.1	12.6	46.1	38.2	36.2	45.8	38.1	37.7	34.2	46.2	29.1	43.1	34.2
Live (CBS)	18.5	27.4	19.9	35.0	16.1	24.3	33.7	8.9	10.8	18.3	16.7	38.9	20.3	23.8	26.2
Live (CTPE)	0.3	24.1	15.0	14.0	5.8	29.6	14.5	26.7	16.7	3.7	39.8	17.2	6.1	8.6	1.9

Table A.69 – Coefficient of Variation for all Segments and all Conditions for Participant 13

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	43.4	36.6	64.5	34.2	64.7	41.1	40.5	65.5	24.8	30.4	77.5	89.9	41.2	65.9	109.3
Live (CBS)	23.6	33.9	45.3	47.9	26.4	45.3	15.0	21.4	60.4	58.6	62.2	59.4	42.6	35.1	61.4
Live (CTPE)	10.3	38.0	16.7	32.3	17.9	25.7	25.3	47.1	52.3	22.6	13.2	29.0	21.5	46.1	24.1

Table A.70 – Coefficient of Variation for all Segments and all Conditions for Participant 14

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	28.3	22.2	36.5	18.6	26.6	17.9	27.5	30.2	26.9	27.8	37.6	32.0	30.9	19.5	14.9
Live (CBS)	8.9	39.9	44.7	51.9	52.9	29.4	8.5	25.4	39.3	26.9	30.7	27.8	49.3	25.7	17.5

Live (CTPE)	7.0	31.8	48.3	16.7	4.5	15.4	37.8	3.2	14.1	36.4	2.5	21.2	21.4	25.2	38.5

Table A.71 – Coefficient of Variation for all Segments and all Conditions for Participant 15

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	22.9	26.8	31.1	43.8	29.8	30.5	15.7	41.4	18.1	55.3	40.8	57.4	72.7	47.5	45.9
Live (CTPE)	14.2	21.2	41.3	12.4	8.0	50.1	63.5	43.6	50.8	48.0	25.5	41.8	8.5	35.9	18.2

Table A.72 – Coefficient of Variation for all Segments and all Conditions for Participant 16

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	10.1	18.6	31.9	23.5	23.9	36.8	21.6	5.9	23.3	96	26.2	32.3	27.6	21.8	44.3
Live (CTPE)	10.9	6.4	5.3	31.7	36.1	35.3	27.6	30.5	23.1	16.6	20.0	41.8	18.3	20.7	26.1

Table A.73 – Coefficient of Variation for all Segments and all Conditions for Participant 17

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	28.9	28.5	24.5	24.2	16.1	33.1	30.1	34.2	63.0	10.2	33.3	37.7	28.0	11.3	37.3
Live (CTPE)	10.8	31.2	47.5	33.3	11.7	8.3	32.6	27.0	39.0	22.3	41.5	34.4	32.2	42.0	33.9

Table A.74 – Coefficient of Variation for all Segments and all Conditions for Participant 18

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	24.2	45.4	61.5	27.4	51.3	41.1	65.0	86.6	66.7	28.9	84.4	50.3	117.3	117.7	68.9
Live (CBS)	12.8	11.4	22.2	17.4	39.1	10.6	4.8	26.9	46.3	49.6	64.9	68.9	23.0	70.7	40.8
Live	42.0	27.5	6.4	35.8	2.0	25.9	31.7	15.8	10.2	32.9	19.5	16.6	4.4	46.4	18.8

(CTPE)

Table A.75 – Coefficient of Variation for all Segments and all Conditions for Participant 19

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	42.4	44.8	63.6	32.6	41.0	32.6	46.3	48.3	53.8	33.2	32.5	28.4	31.8	12.1	36.3
Live (CBS)	9.6	5.8	17.7	5.5	22.7	26.5	29.3	16.9	15.0	17.1	11.9	27.5	31.0	20.2	33.0
Live (CTPE)	27.2	28.6	32.1	8.1	20.2	6.5	21.7	56.0	14.8	34.1	49.6	18.2	24.5	45.5	69.5

Table A.76 – Coefficient of Variation for all Segments and all Conditions for Participant 20

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	18.6	25.5	44.2	43.9	38.2	39.2	26.8	37.7	40.5	50.6	51.0	37.5	34.5	37.5	52.2

Live (CBS)	25.3	33.9	21.6	15.4	17.9	31.2	29.3	49.9	20.8	13.7	50.9	18.1	9.5	26.4	16.2
Live (CTPE)	9.2	27.1	10.5	24.1	20.0	14.7	32.4	8.0	9.3	23.9	8.0	24.5	20.7	12.9	24.6

Table A.77 – Coefficient of Variation for all Segments and all Conditions for Participant 21

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	46.1	39.3	41.6	24.1	29.0	28.3	32.0	44.5	34.7	68.0	37.1	10.4	48.5	39.1	39.7
Live (CBS)	9.5	9.8	19.9	12.4	12.5	31.7	3.9	30.5	34.5	5.0	11.0	29.7	28.2	21.3	29.4
Live (CTPE)	29.0	14.1	19.8	33.6	18.2	11.1	18.3	16.3	26.2	6.6	33.3	30.0	28.8	16.1	17.0

Table A.78 – Coefficient of Variation for all Segments and all Conditions for Participant 22

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**

Machine	23.9	41.9	61.3	33.9	47.7	49.6	40.9	27.4	40.8	8.3	31.9	63.8	56.7	50.1	67.2
Live (CBS)	14.6	15.6	11.5	11.9	28.4	27.5	27.4	27.6	11.0	13.0	23.2	29.6	12.1	20.9	23.6
Live (CTPE)	13.8	28.8	41.5	6.5	14.4	23.7	31.2	18.7	16.1	56.5	39.8	48.2	26.5	29.9	21.0

Table A.79 – Coefficient of Variation for all Segments and all Conditions for Participant 23

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Machine	47.6	31.7	13.8	34.0	10.3	26.9	40.2	20.7	21.4	32.4	46.3	31.9	27.9	29.4	56.0
Live (CBS)	54.2	16.0	36.1	32.2	43.8	40.1	48.9	61.5	56.8	33.6	49.6	46.6	5.0	18.2	8.4
Live (CTPE)	11.4	10.5	35.5	32.4	21.4	19.3	30.9	44.3	16.4	34.1	24.0	42.6	10.3	44.7	37.7

Table A.80 – Coefficient of Variation for all Segments and all Conditions for Participant 24

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		

Condition	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot
Live (CBS)	15.0	63.6	63.9	23.6	60.7	34.9	32.1	32.4	32.0	70.6	47.9	59.9	55.1	39.1	68.0
Live (CTPE)	13.5	35.0	4.5	35.3	9.1	28.3	11.8	15.9	31.2	28.5	15.7	31.4	14.5	16.4	11.0

Table A.81 – Coefficient of Variation for all Segments and all Conditions for Participant 25

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	22.2	22.5	23.4	6.6	48.2	35.4	6.1	22.5	5.3	23.4	7.2	14.2	13.6	28.5	40.3
Live (CTPE)	28.5	23.1	65.3	13.1	38.5	4.3	13.6	17.2	34.1	47.1	29.5	7.4	39.0	57.5	63.3

Table A.82 – Coefficient of Variation for all Segments and all Conditions for Participant 26

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**

Live (CBS)	17.3	26.2	11.0	32.1	10.4	15.0	14.3	27.3	10.4	33.2	33.9	31.7	2.4	32.1	37.3
Live (CTPE)	34.5	30.9	10.5	34.1	19.2	17.4	1.5	15.6	8.4	13.6	33.2	30.0	11.4	16.6	15.8

Table A.83 – Coefficient of Variation for all Segments and all Conditions for Participant 27

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	16.4	22.0	19.4	12.3	50.4	27.9	5.8	12.2	44.4	27.0	45.2	26.9	4.2	16.9	32.2
Live (CTPE)	42.3	9.4	36.4	34.1	21.1	21.9	11.4	24.8	6.2	36.4	46.9	9.7	16.6	32.2	16.6

Table A.84 – Coefficient of Variation for all Segments and all Conditions for Participant 28

	Cervical			**Upper Thoracic**			**Lower Thoracic**			**Upper Lumbar**			**Lower Lumbar**		

Condition	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot	FE	LB	Rot
Live (CBS)	30.9	34.8	12.2	34.9	11.3	28.1	40.6	30.5	35.7	29.0	15.2	12.1	29.7	31.5	34.5
Live (CTPE)	18.8	37.2	22.5	22.4	15.2	16.6	27.4	21.0	53.5	27.0	26.7	47.2	9.1	18.7	21.3

Table A.85 – Coefficient of Variation for all Segments and all Conditions for Participant 29

	Cervical			Upper Thoracic			Lower Thoracic			Upper Lumbar			Lower Lumbar		
Condition	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**	**FE**	**LB**	**Rot**
Live (CBS)	7.5	20.6	28.4	23.3	35.7	10.5	13.8	22.3	20.2	26.9	16.3	5.7	27.1	15.4	30.5
Live (CTPE)	33.0	28.6	42.8	17.5	8.9	26.0	12.4	17.3	27.5	34.9	21.4	38.8	16.9	16.4	19.6

A.1.5.6. Force-EMG Data

Correlation Curves

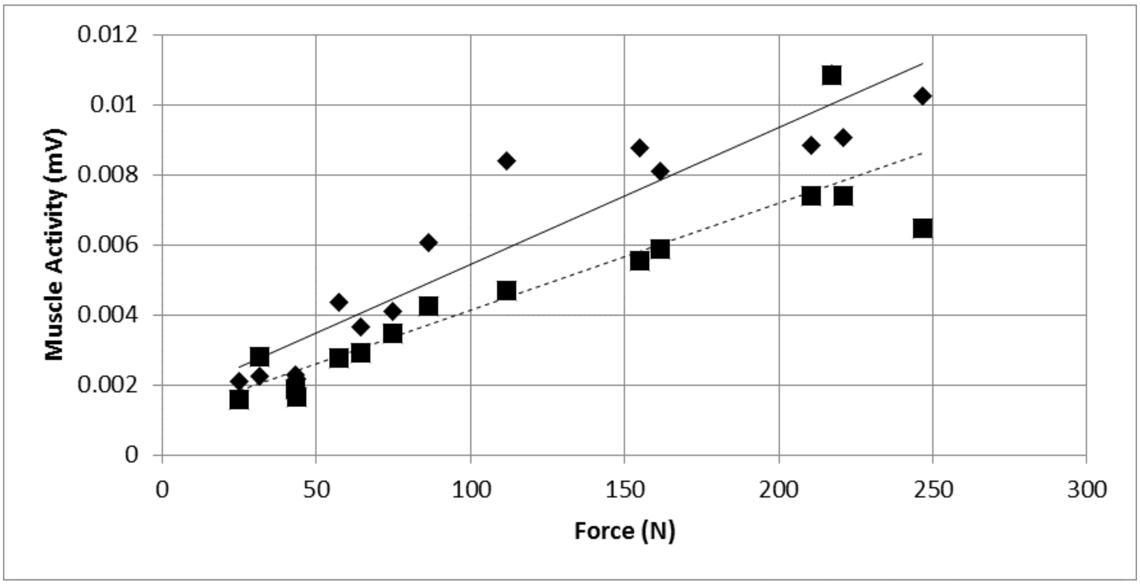


Figure A.27 – Force-EMG correlation curve from three gradually increasing exertion trials participant 9. Diamonds – Left Side; Square – Right Side. Solid line – Left side trend line, R^2=0.8933; Dashed line – Right side trend line, R^2=0.8296

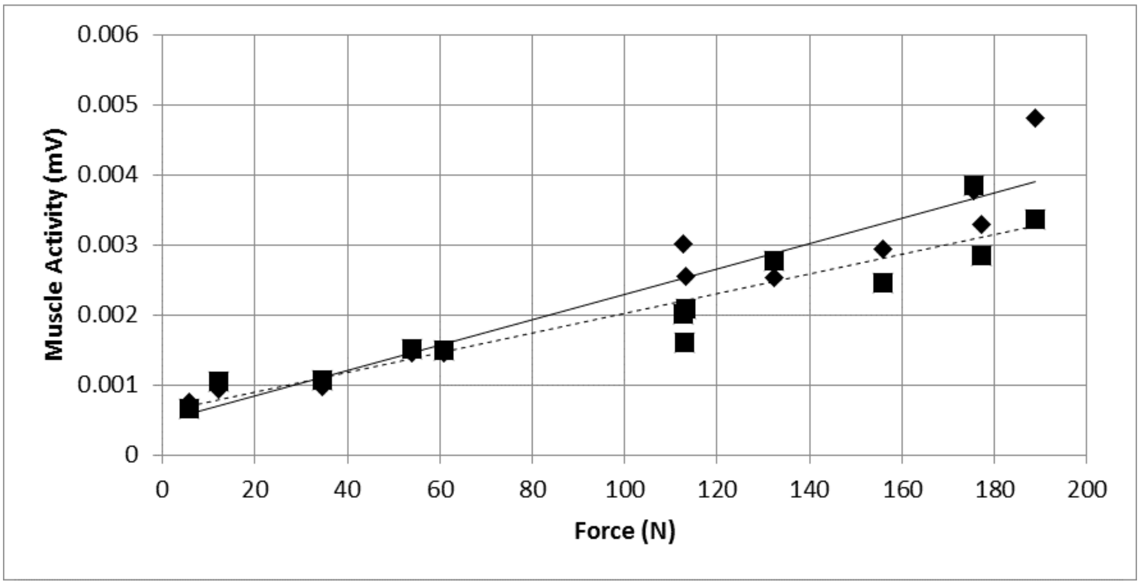


Figure A.28 – Force-EMG correlation curve from three gradually increasing exertion trials participant 10. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.8974; Dashed line – Right side trend line, R^2=0.8739

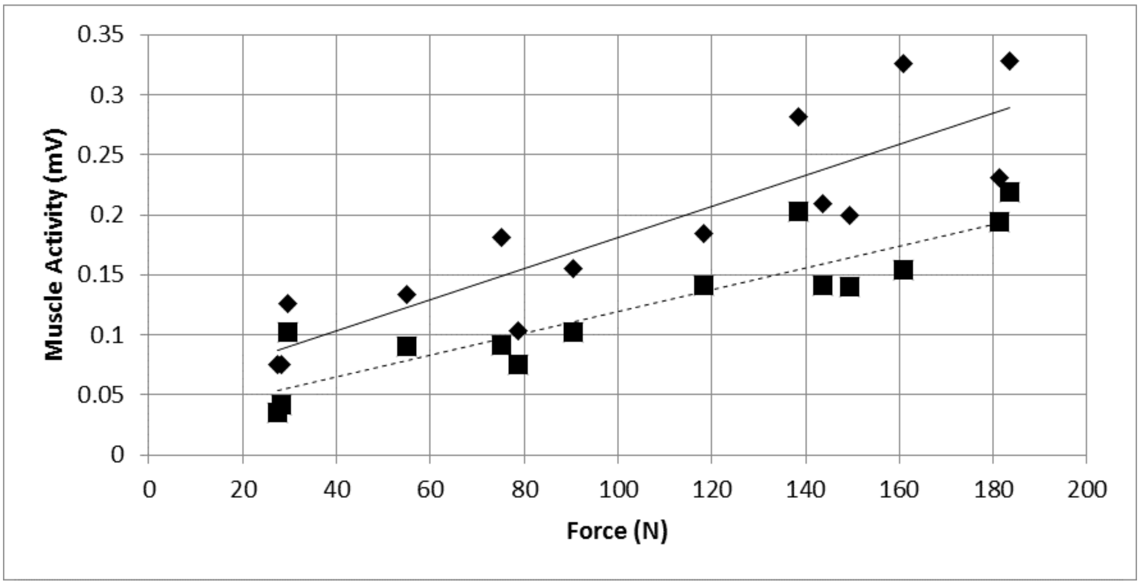


Figure A.29 – Force-EMG correlation curve from three gradually increasing exertion trials participant 11. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.7739; Dashed line – Right side trend line, R^2=0.8122

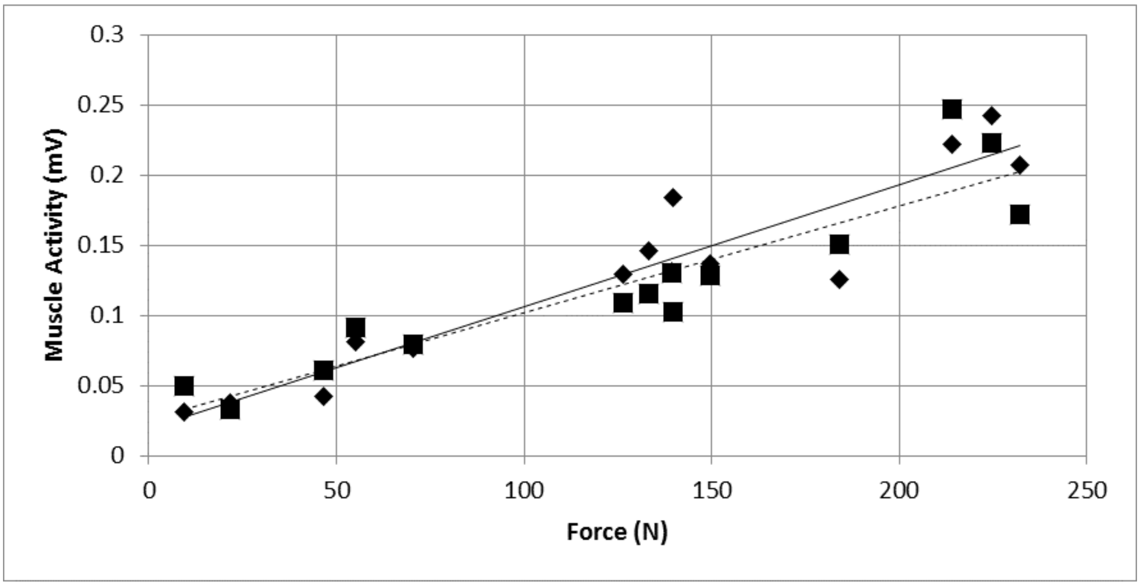


Figure A.30 – Force-EMG correlation curve from three gradually increasing exertion trials participant 12. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.8875; Dashed line – Right side trend line, R^2=0.8484

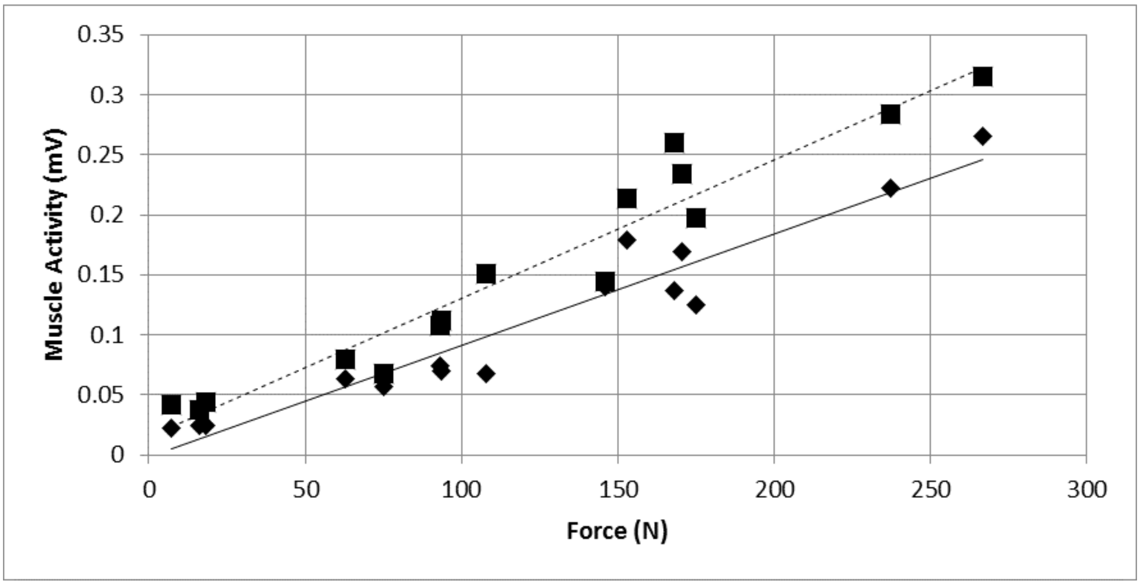


Figure A.31 – Force-EMG correlation curve from three gradually increasing exertion trials participant 14. Diamonds – Left Side; Square – Right Side. Solid line – Left side trend line, R^2=0.9369; Dashed line – Right side trend line, R^2=0.9289

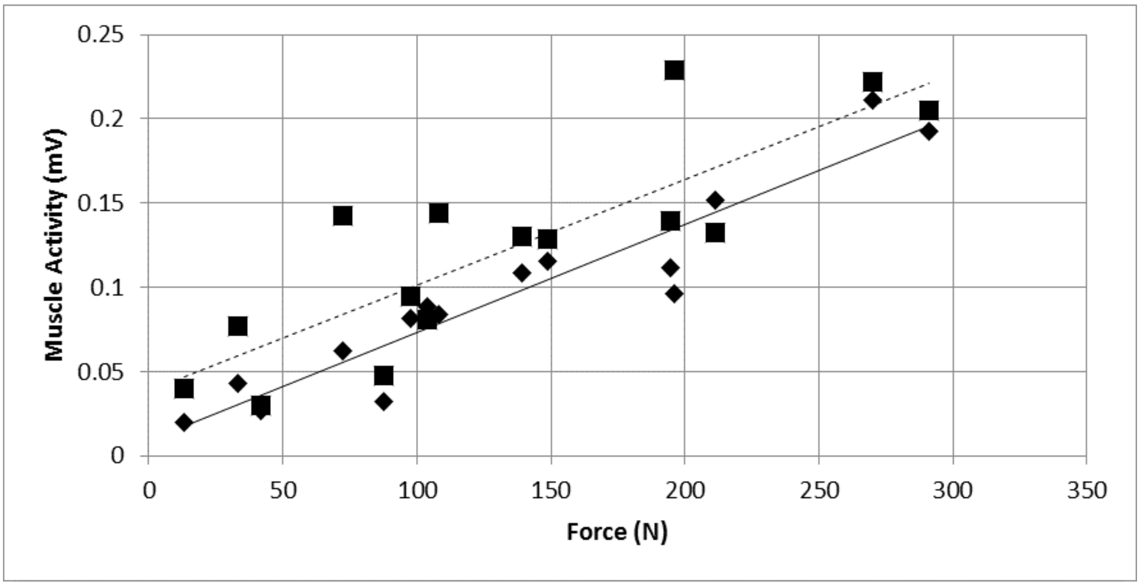


Figure A.32 – Force-EMG correlation curve from three gradually increasing exertion trials participant 19. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.7069; Dashed line – Right side trend line, R^2=0.8942

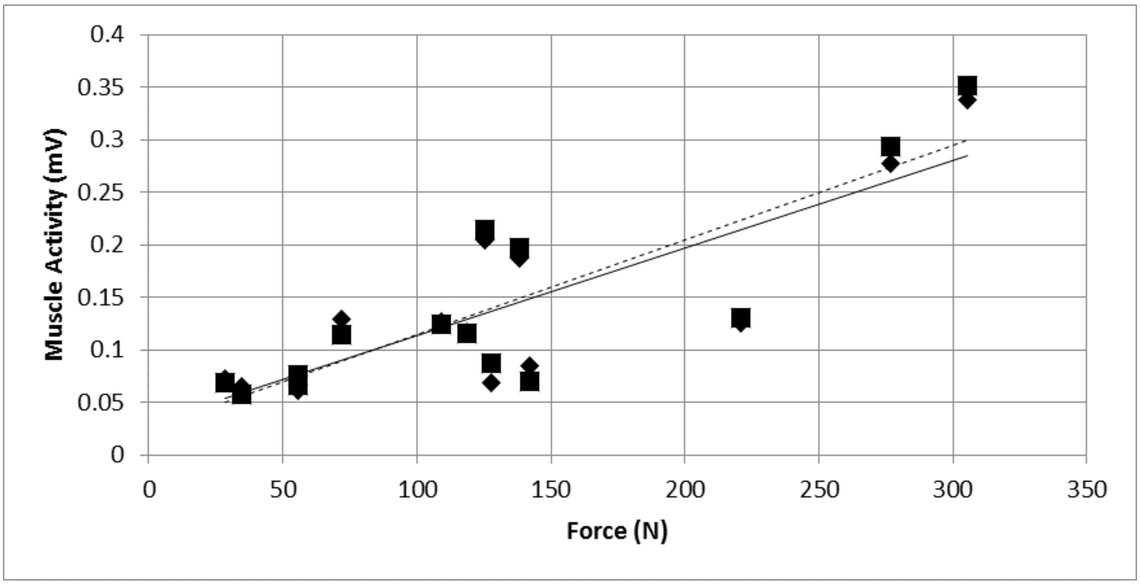


Figure A.33 – Force-EMG correlation curve from three gradually increasing exertion trials participant 20. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.6965; Dashed line – Right side trend line, R^2=0.7135

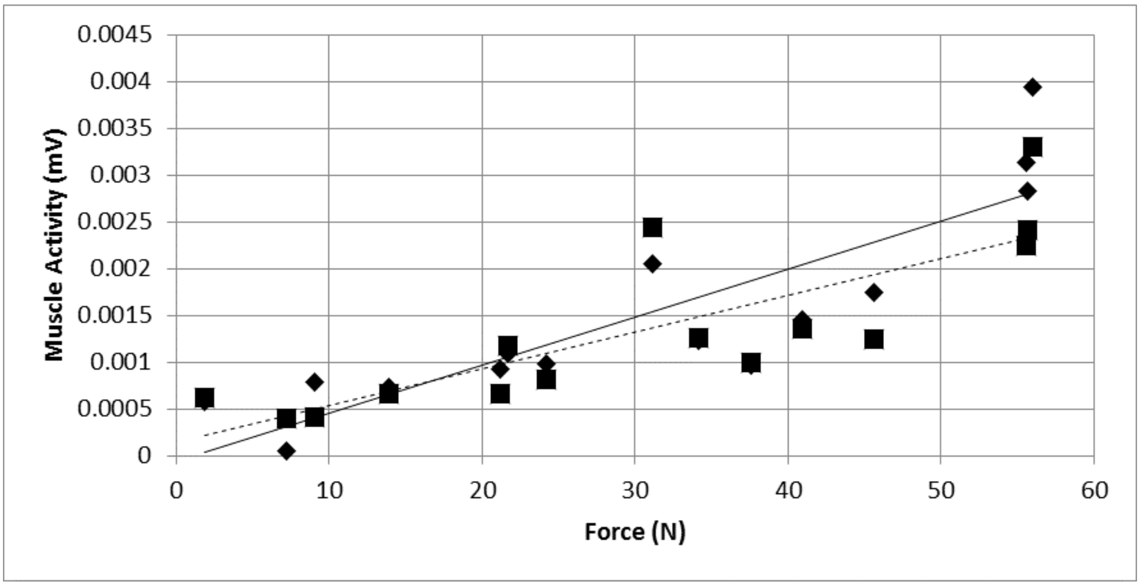


Figure A.34 – Force-EMG correlation curve from three gradually increasing exertion trials participant 22. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.7572; Dashed line – Right side trend line, R^2=0.6716

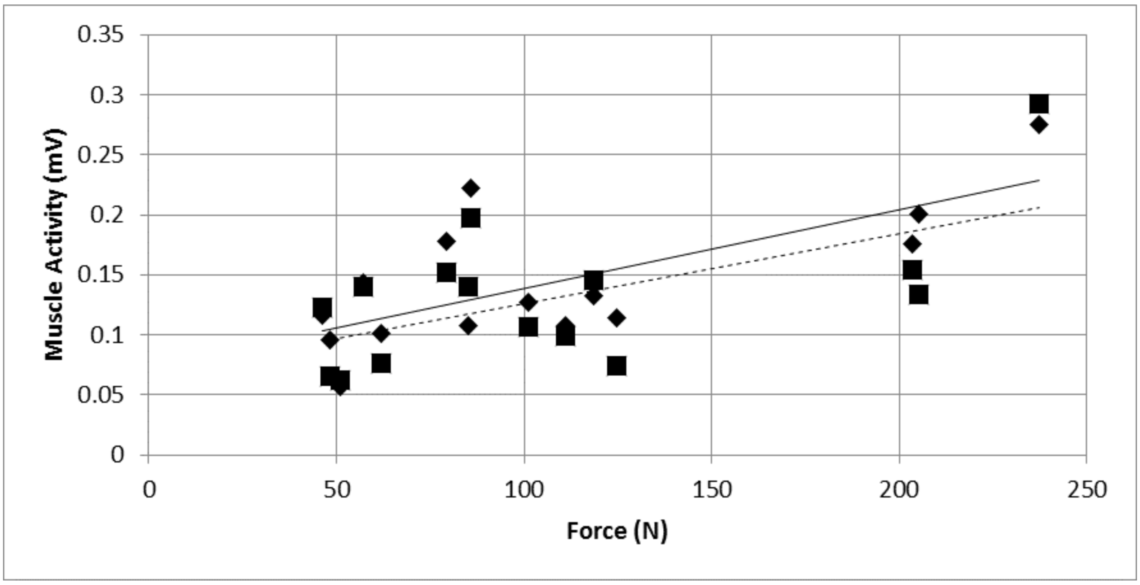


Figure A.35 – Force-EMG correlation curve from three gradually increasing exertion trials for participant 23. Diamonds – Left Side; Square – Right Side Solid line – Left side trend line, R^2=0.3665; Dashed line – Right side trend line, R^2=0.5008

Machine and Live Scrummaging Data

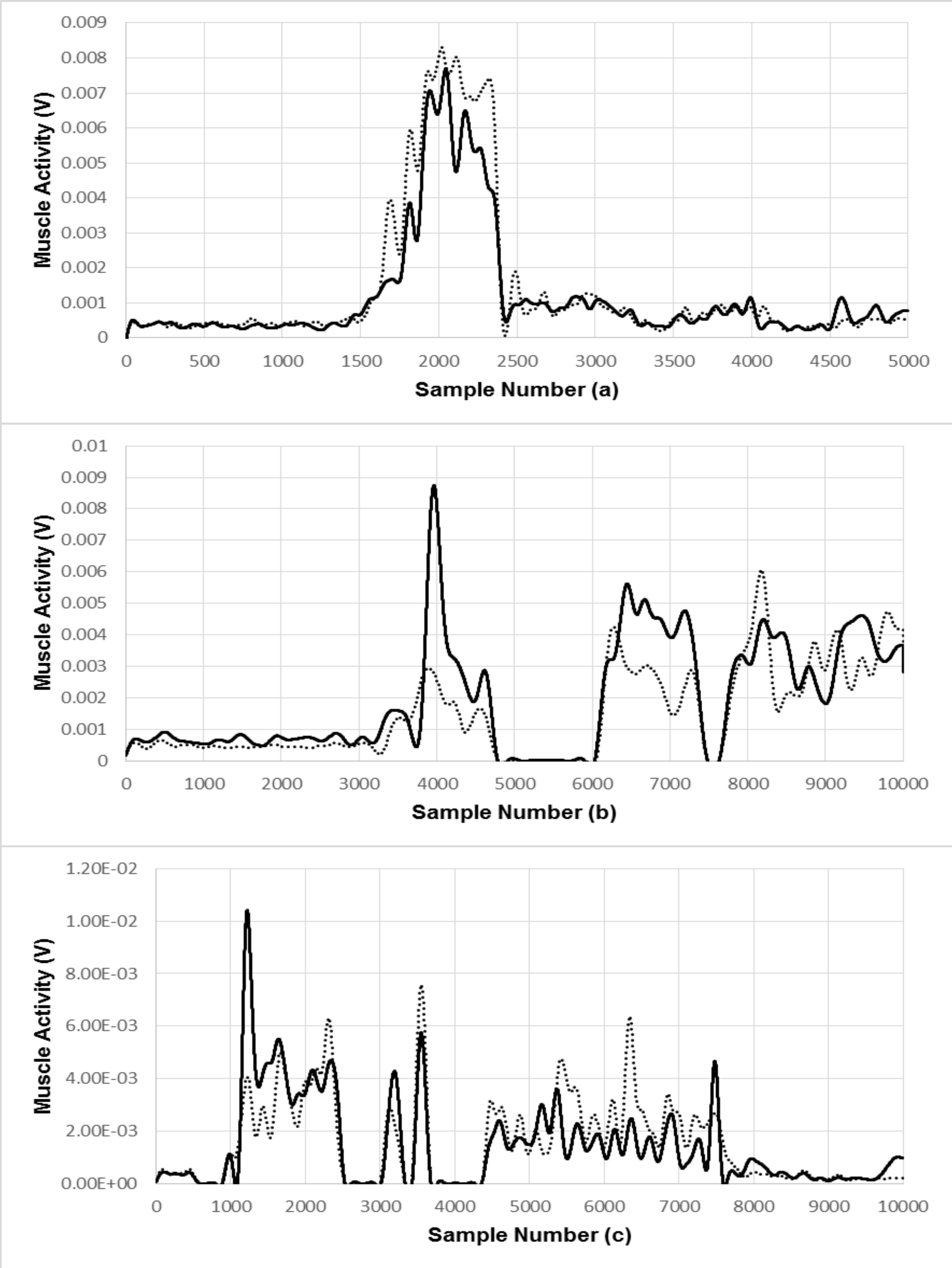


Figure 36 - Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 9. Solid line - Right CES; Dotted line - Left CES

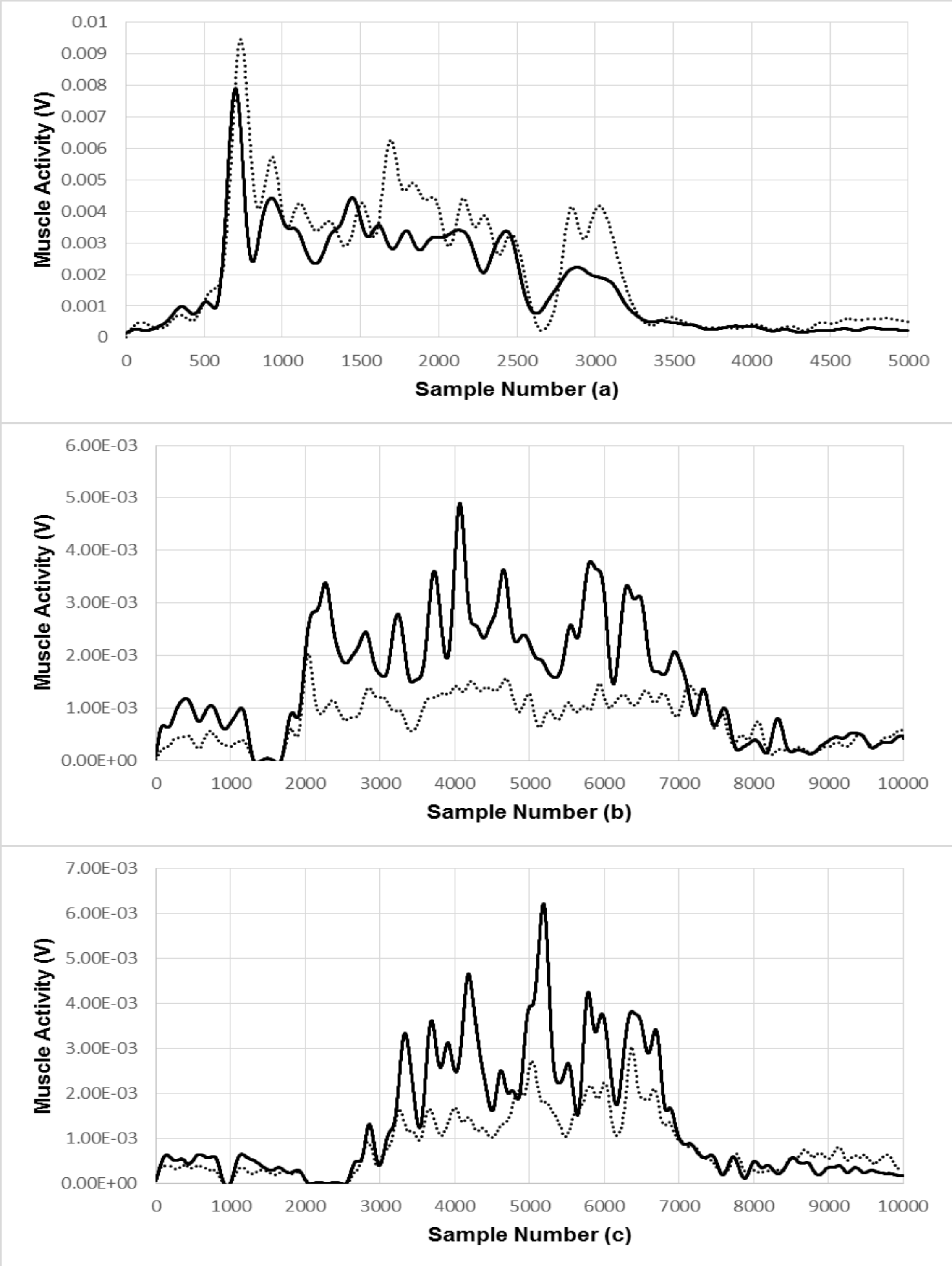


Figure A.37 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 10. Solid line - Right CES; Dotted line - Left CES

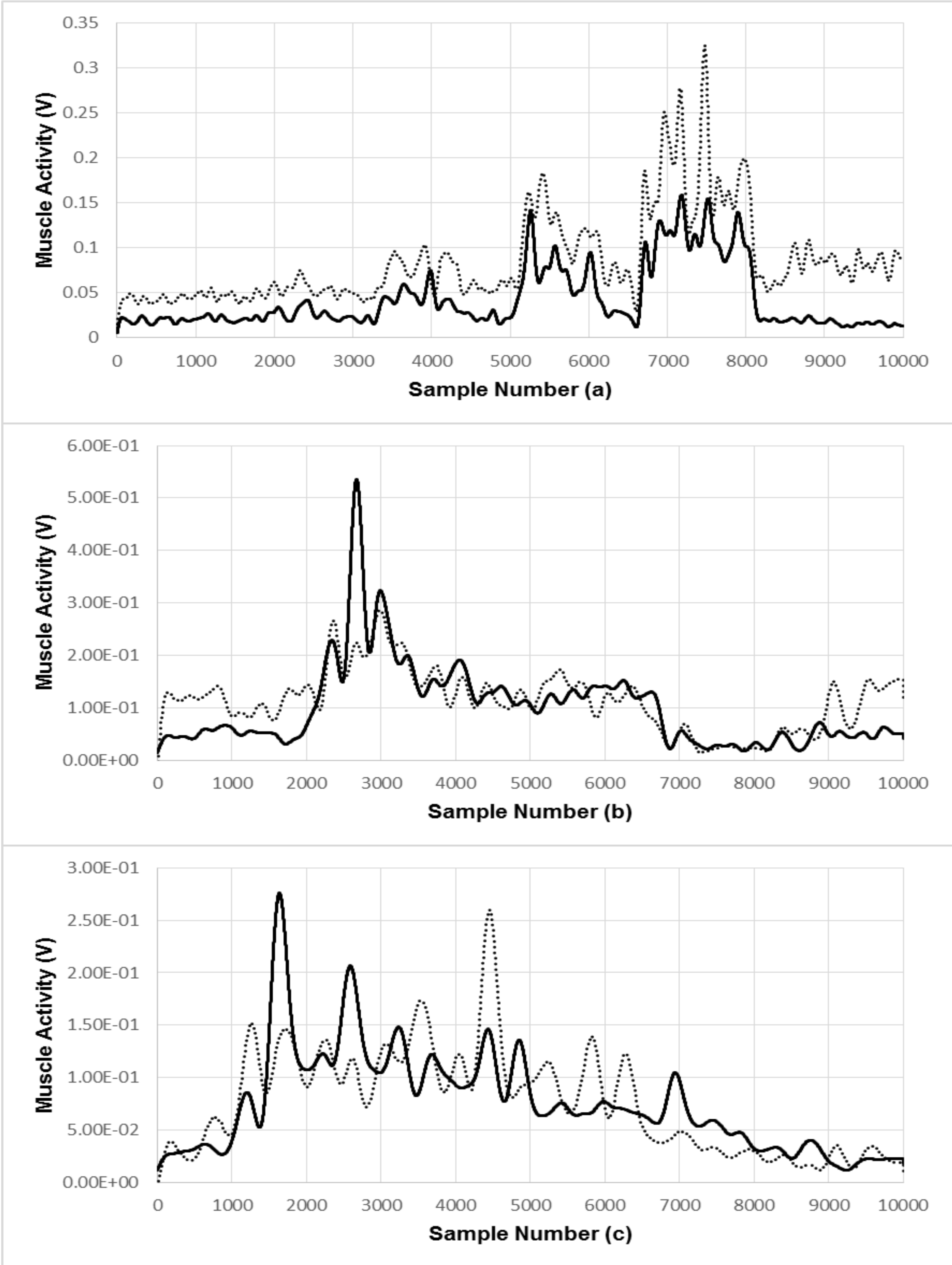


Figure A.38 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 11. Solid line - Right CES; Dotted line - Left CES

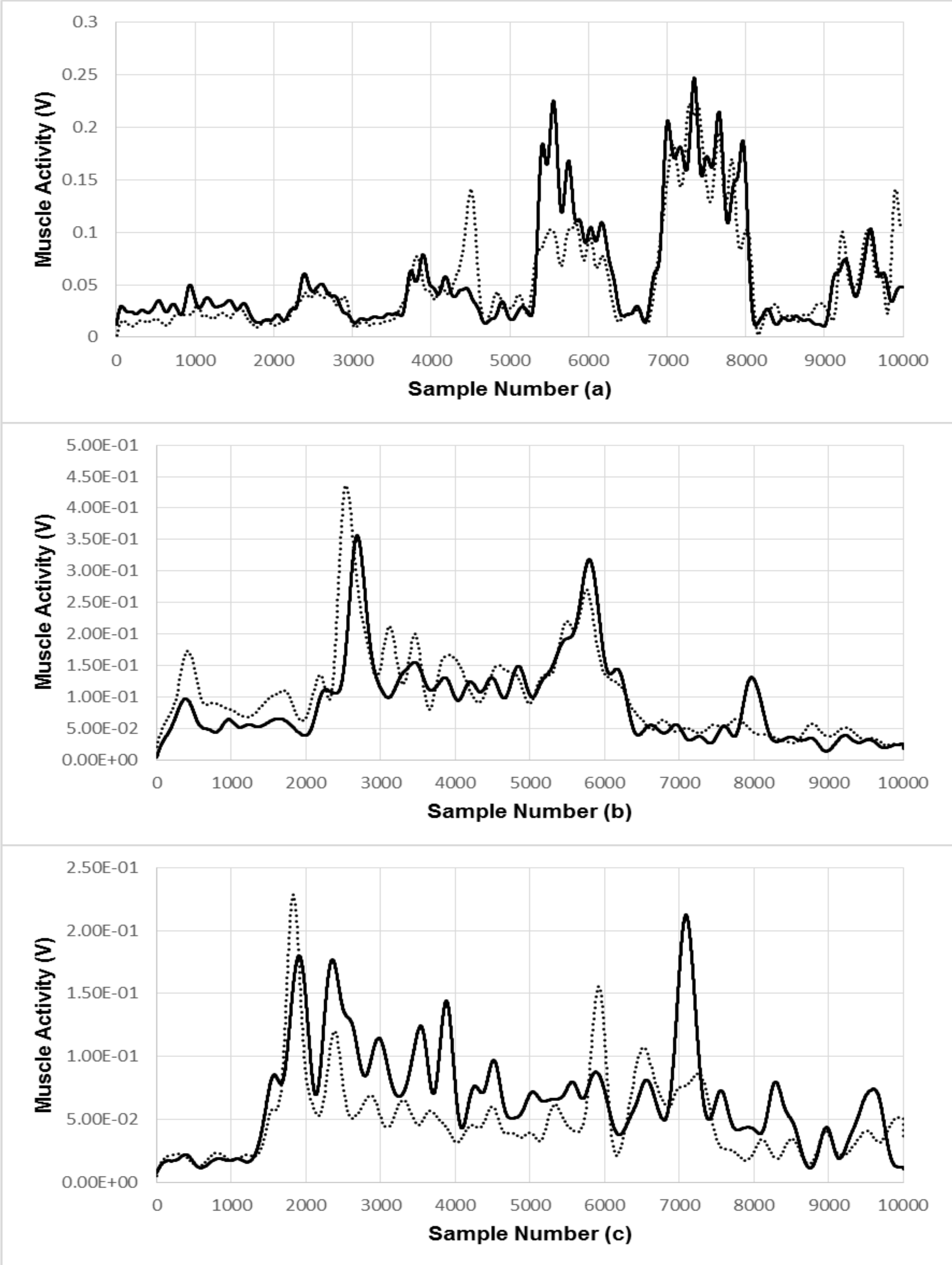


Figure A.39 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 12. Solid line - Right CES; Dotted line - Left CES

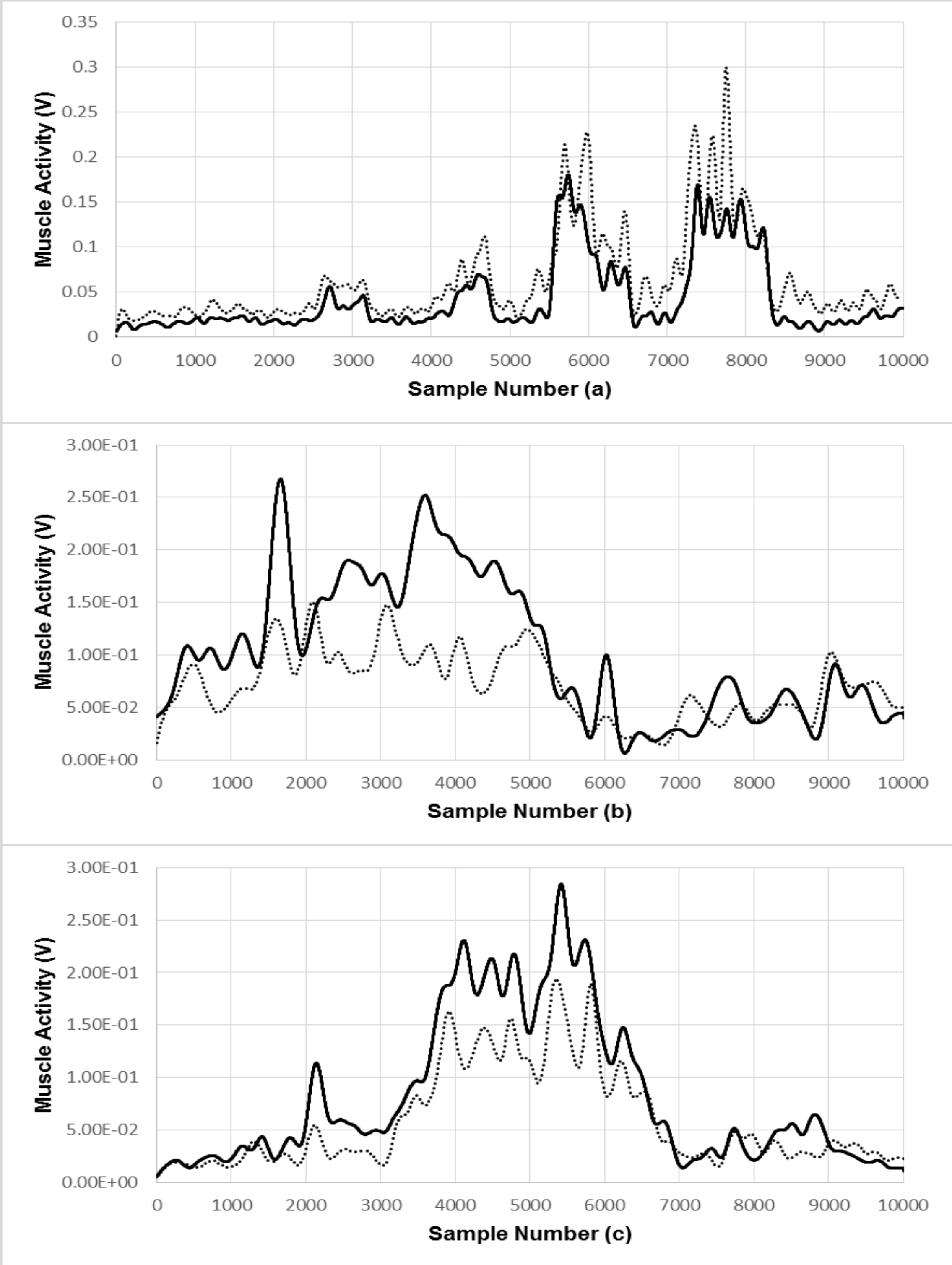


Figure A.40 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 14. Solid line - Right CES; Dotted line - Left CES

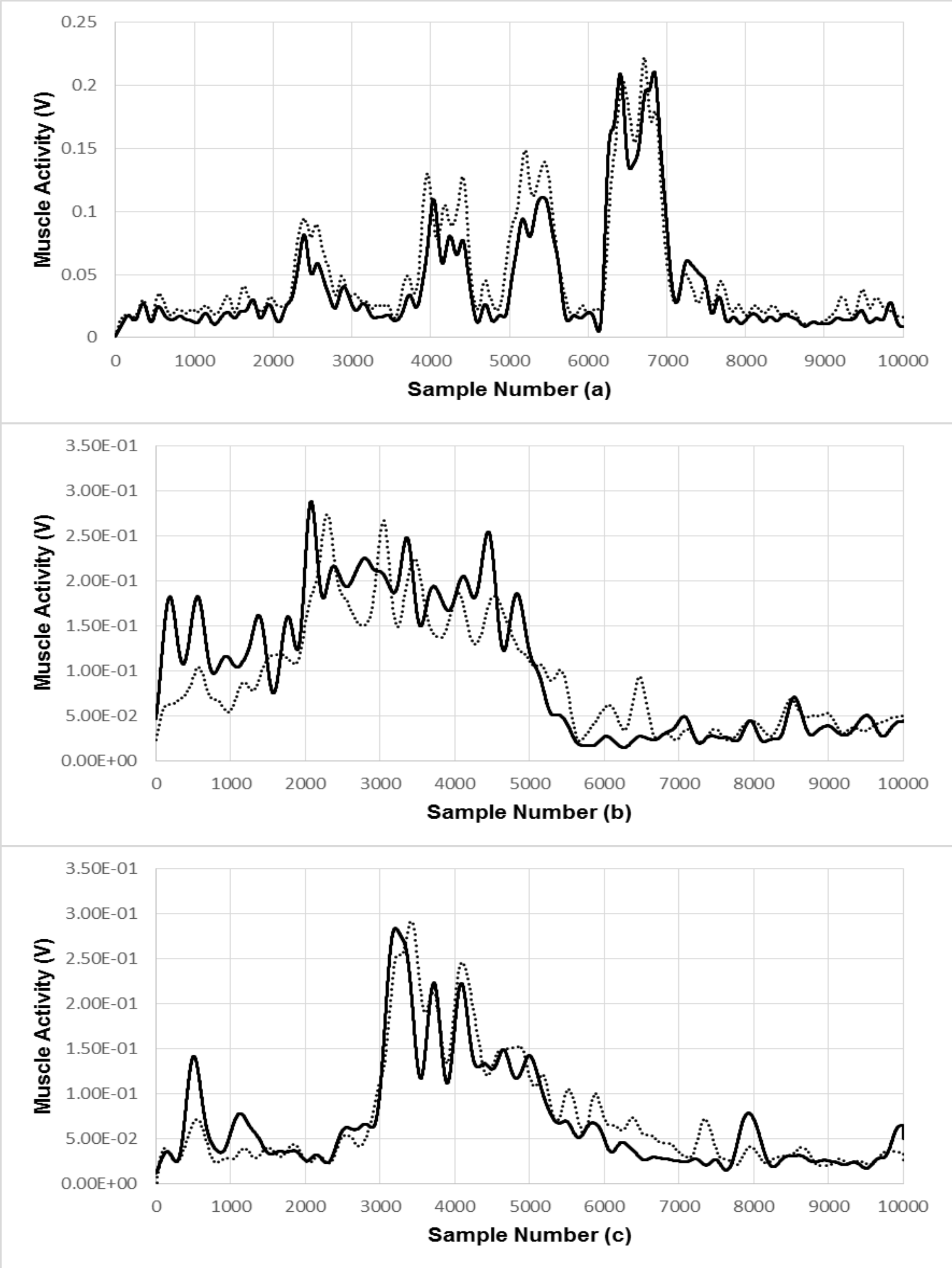


Figure A.41 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 19. Solid line - Right CES; Dotted line - Left CES

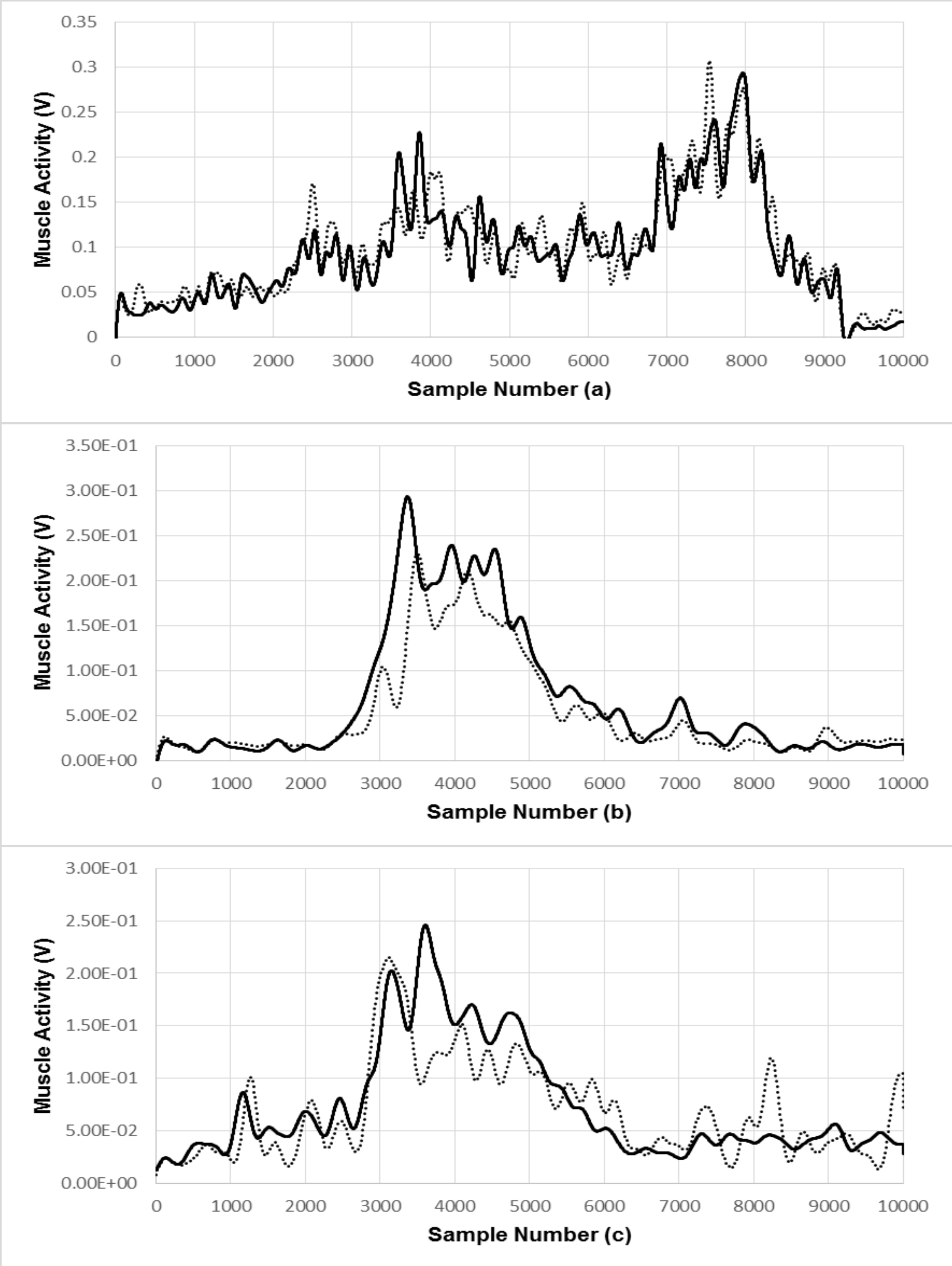


Figure A.42 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 20. Solid line - Right CES; Dotted line - Left CES

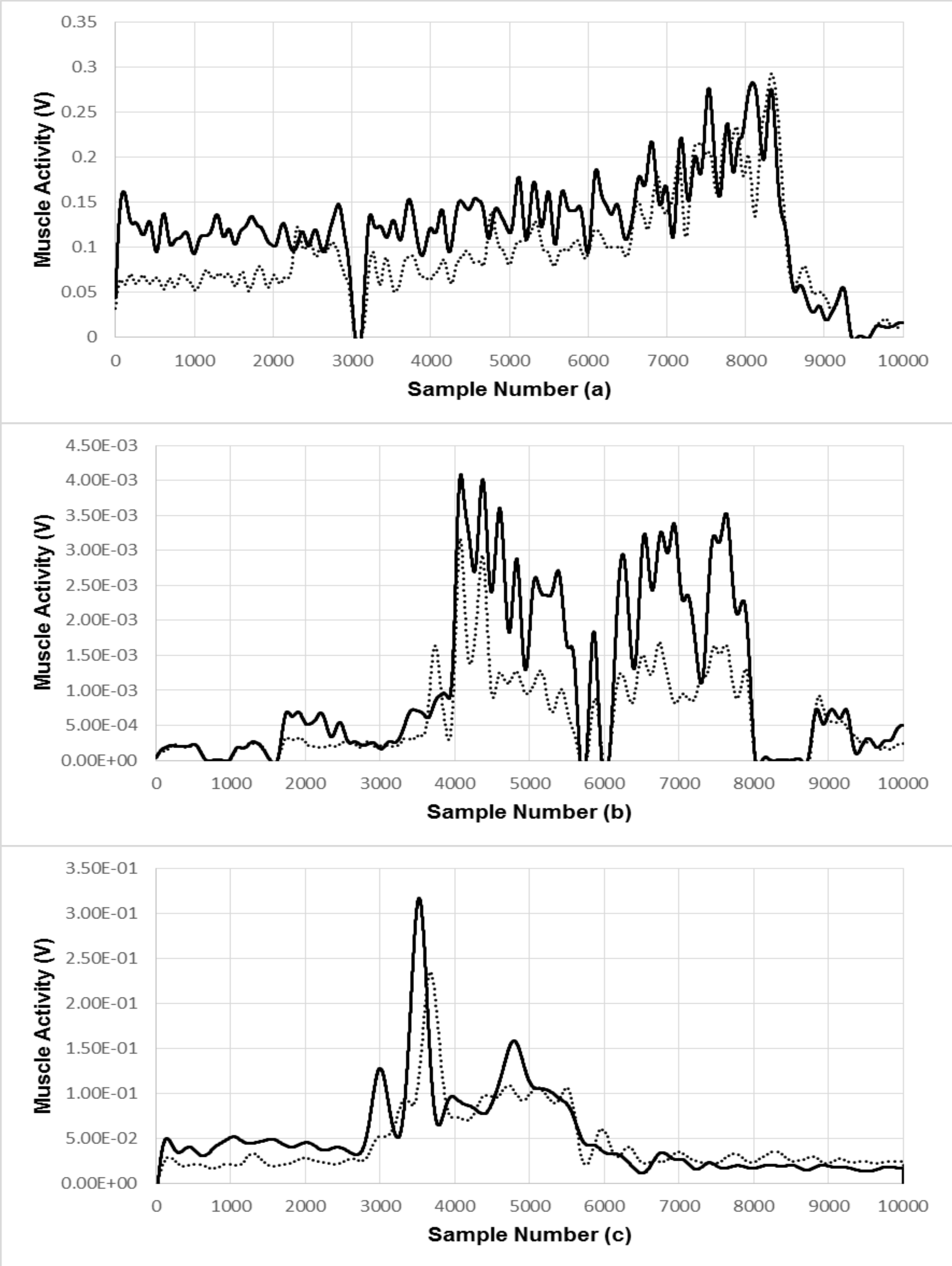


Figure A.43 – Example EMG traces of the left and right CES muscle during (a) machine, (b) live (CBS), and (c) live (CTPE) scrummaging for Participant 23. Solid line - Right CES; Dotted line - Left CES

A.1.6. Participant Consent Form

PARTICIPANT CONSENT FORM

An investigation into spinal kinematics and neck loading, through the use of electromyography, during rugby scrummaging

Please initial as appropriate

1. I can confirm that I have read and understood the Participant Information Sheet for the above study (Project ID: SCR001) and I have had the opportunity to ask questions and have had them answered to my satisfaction. ☐

2. I can confirm that I have completed the Participant Details Form for the above study (Study Number: SCR001). ☐

3. I can confirm that I have been instructed in and understand the testing procedure for the above study (Study Number: SCR001). I feel I have a full understanding of what is required of me and have had any questions I have answered to my satisfaction. ☐

4. I understand that I am being observed and that my experimental data will be used for research purposes only. I can further confirm that I understand my details will not be used to identify experimental data when disseminated to the public domain. ☐

5. I understand that my participation is voluntary and I am free to withdraw at any time, without giving any reason. ☐

6. I agree to participate in the above study. ☐

____________________________ _______________ _______________

Participant Name (BLOCK CAPITALS) Signature Date

____________________________ _______________ _______________

Researcher Name (BLOCK CAPITALS) Signature Date

A.1.7. Participant Details Forms

A.1.7.1. Version 1

PARTICIPANT DETAILS FORM
An investigation into spinal kinematics and neck loading, through the use of electromyography, during rugby scrummaging

The information below is recorded to provide an appreciation of the overall demographic of the study. In no way will it be used to identify experimental data or used for any other purpose other than that which was outlined in the Participant Information Sheet. On the Participant Consent Form, you will be asked if you have completed this form to allow data anonymisation.

Please fill as appropriate:

1. Age:
2. Height: _______m_____cm OR _______feet______inches
3. Weight: _______kg OR _______stone_____pounds
4. Playing position (please tick all that apply):
   - ☐ Hooker
   - ☐ Loose-head prop
   - ☐ Tight-head prop
5. Playing experience in front-row (years): _____________
6. Do you take part in any other sports on a regular basis other than rugby?
   - ☐ Yes
   - ☐ No
7. If yes, please specify: ____________________________
8. Are you left or right handed?
   - ☐ Left
   - ☐ Right
9. Have you ever had or do you currently have any head, neck or spinal injuries?
   - ☐ Yes
   - ☐ No
10. If yes, please specify what sort of injury it was (fracture, ligament, muscle strain etc.):
____
____
____
11. Neck Circumference: _______________cm
12. Shoulder Circumference: ____________cm
13. Chest Circumference: _______________cm

A.1.7.2. Version 2

PARTICIPANT DETAILS FORM
An investigation into spinal kinematics and neck loading, through the use of electromyography, during rugby scrummaging

The information below is recorded to provide an appreciation of the overall demographic of the study. In no way will it be used to identify experimental data or used for any other purpose other than that which was outlined in the Participant Information Sheet. On the Participant Consent Form, you will be asked if you have completed this form to allow data anonymisation.

Please fill as appropriate:

1. Age:
2. Height: _______m_____cm OR _______feet___ inches
3. Weight: __________kg OR _______stone__ pounds
4. Cautions/contraindications

Allergies	yes [ ] no [ ]
Systemic steroids	yes [ ] no [ ]
Anticoagulants / blood clotting disorders	yes [ ] no [ ]
History of any seizures	yes [ ] no [ ]
Diabetes	yes [ ] no [ ]
Low blood pressure	yes [ ] no [ ]
Exhibiting uncontrolled movements	yes [ ] no [ ]
Patient confused	yes [ ] no [ ]

Particular notes:

____

____

5. Playing position (please tick all that apply):
   - ☐ Hooker
   - ☐ Loose-head prop
   - ☐ Tight-head prop
6. Playing experience in front-row (years): _________________
7. What level do you play at? _____________________________

8. Playing information

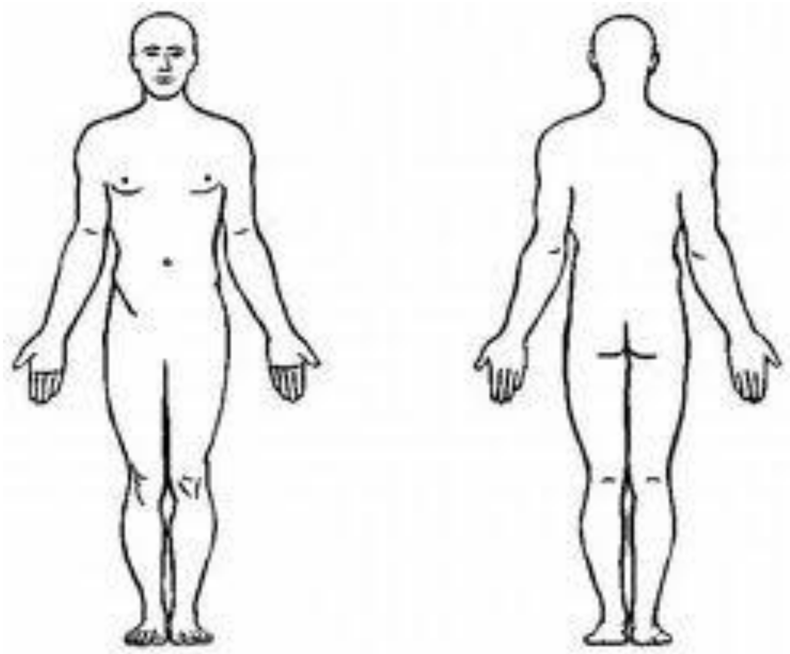
Training sessions per week?	
Number of scrummages per session?	
Do you wear headgear?	
What surface(s) do you scrummage on?	

9. Other activities

What other sports do you play? (please specify)	
How many times per week do you exercise?	
What other physical activities do you do? (eg. rock climbing)	

10. Are you left or right handed?
    - ☐ Left
    - ☐ Right

11. On the image below please detail any injuries you have had in the current playing season and what types of injuries they were in the space below (sprain, fracture, concussion etc.)



____
____
____

12. Please note any other serious physical injuries, operations or medical conditions that you have which may affect spinal movement.

____

13. Neck Circumference: ___________________cm

14. Shoulder Circumference: ________________cm

15. Chest Circumference: ___________________cm

A.1.8. Participant Information Sheet

PARTICIPANT INFORMATION SHEET

An investigation into spinal kinematics and neck loading, through the use of electromyography, during rugby scrummaging

1. Study title

An investigation into spinal motion and forces exerted on the neck during rugby scrummaging.

2. Invitation paragraph

You have been invited to participate in a research study carried out by the Institute of Medical Engineering and Medical Physics (IMEMP), Cardiff University. Before you decide whether or not to participate it is important for you to understand the reasons why the research is being carried out and what participation involves. Please take time to read the following information and discuss it with others if you wish. Ask the Research Team if you have any questions or if you require further information. Contact details are available at the end of this document. Please take your time deciding whether or not you wish to participate.

Before you participate, the investigators will give you an opportunity to ask any questions you may have.

Thank you for taking the time to read this.

3. What is the purpose of the study?

A number of studies have looked at the forces generated in rugby scrummaging against a scrum machine however there has not been any research into the level of force the neck is subjected to during live scrummaging. The reason this is of interest is that neck injuries are widely reported in the literature but there is not a full understanding of the injury mechanisms involved. This study aims to determine the forces generated in the neck of a front row player through the use of electromyography (EMG). EMG monitors the level of electrical activity when a muscle contracts; the greater the muscle contraction (and therefore force exerted), the greater the electrical activity. Thus, by monitoring the electrical activity of the main extensor muscles of the neck, the level of force resisted by the neck can be determined. As well as this aspect of the study, spinal kinematics will also be monitored. Kinematics is the study of motion and thus incorporates distances travelled, speeds, and accelerations of a specific point. Often these values are given relative to another point. Inertial sensors will be placed on the spine to investigate the kinematics of specific vertebra during scrummaging. Data will be acquired during regular rugby training sessions. The study will last approximately 6 months.

4. Why have I been chosen?

You have been chosen for this study as you are currently a front row player actively taking part in rugby and rugby-specific training. You have been trained to play in your specific position and thus know the technique of scrummaging which is of particular importance for this study. In this study there will be 20-25 participants that have front row playing experience. The number of participants may expand depending on the results.

5. Do I have to take part?

You have no obligation to take part and participation is entirely voluntary. If you do decide to take part you will be required to sign a Participant Consent Form and fill in a Participant Details Form. You are free to withdraw from the study at any time without giving a reason.

6. What will happen to me if I take part?

The setup and trials should last no longer than 30 minutes. The whole research project will last approximately 6 months. This means that all participants will be seen in this time period providing they are available during this time. Depending on the results, you may be asked to return for a repeat trial. However, this is unlikely and you are always free to withdraw from the study as mentioned previously.

7. What about confidentiality?

Prior to the findings being disseminated in the public domain, all direct and indirect forms of identification will be removed through the anonymisation of data to provide confidential data handling. All data will be anonymised by separating Participant Details Forms, Participant Consent Forms, and the results. All data from the EMG and inertial sensors will be electronic and as such will be stored on a password protected University owned laptop as well as on a password protected University owned desktop computer. All paper parts of the data collection will be securely stored and access to them will only be allowed to the Research Team. All information and data will be treated as strictly confidential.

8. What do I have to do?

Before applying any instrumentation to your body, you will be fully briefed on the entire study, allowing time for any questions you may have. Once you are satisfied, and before testing starts, you will be asked to complete Participant Details and Participant Consent Forms. If you are still willing to participate you will be instrumented with both surface EMG electrodes and inertial sensors.

A string of 6 inertial sensors will be attached directly to your skin at specific spinal landmarks. These will be on your head and spine. There will be one sensor on your forehead, one on your lower neck, two on the middle portion of your back and two on your lower back. The sensors will be attached using double-sided hypoallergenic tape and so there should be no problems with any allergic reactions. However, there may be slight skin irritation when removing the tape after testing, similar to when removing a plaster.

There will also be 6 surface EMG electrodes which are self-adhesive. Two electrodes will be attached, one above the other, roughly 1cm horizontally away from the vertical midline of your neck. This will be done for both the left and right sides so a total of 4 electrodes are attached to the back of your neck. The other 2 electrodes will be placed on your left and right collar bones (clavicle). These two electrodes act as reference electrodes since bony parts of the body are relatively electrically inactive. Therefore the electrical activity detected by the other four electrodes on your neck muscles can be compared back to the reference values to obtain a voltage reading.

Once you are fully instrumented you will be first asked to perform a series of movements to determine your overall spinal and neck range of motion (ROM). These will be done in a standing position and then neck ROM will be done in a scrummaging position. The movements are in a predefined order and this will be made clear before starting the testing. ROM is of interest as literature has shown that rugby players, the front row in particular, have a reduced neck ROM compared to the rest of the population. After this you will be asked to perform a series of isometric muscle contractions of the neck against an instrumented scrum machine. This will also be done in a scrummaging position, and the EMG taken will be of the extensor muscles of the neck. Isometric contractions are when your muscle is contracting but when no actual movement occurs. The contractions will be resisted by an inverted force platform which is a piece of instrumentation that produces a voltage when mechanical load is applied to it. You will push vertically upwards with your head with gradually increasing exertion against the force platform for a period of 10-15 seconds up to your maximum voluntary contraction (MVC). There will be 5 different exertions: minimum, 25%, 50%, 75%, and maximum and you will be asked to pause for a second at each exertion. It does not matter if you are not sure what 25% of your maximum force is. Whatever you perceive to be 25% at the time will be sufficient for the study. Between these trials you will be given 2 minutes rest to allow for full physiological recovery. The increasing force trials will be performed three times in total. For the last machine scrummaging trial, you will be asked to engage as you normally would when training on the scrum machine to determine muscle activity, vertical force exerted by the neck and spinal motion data.

Once this has been done, you will be asked to scrummage as part of a full, live training scrum to determine the level of electrical muscle activity in the extensor muscles of your neck. For the live scrums there will be 3 trials using the old engagement sequence of 'crouch-touch-set' and then 3 trials using the new engagement sequence of 'crouch-bind-set'. This is so that we can determine what the effect of the new law is on the front row. After every trial, you will always be given two minutes to recover. This data will then be equated to a force by using the data collected previously from the force platform and EMG.

9. Are there any risks?

Since you are asked to perform an MVC the risk of injury is extremely low. This is because the contraction is voluntary and you will not be asked to exert yourself to a greater extent than is comfortable for you. It is important to note that scrummaging has inherent risks and this study is not perceived to increase the risk of injury in any way.

Although the risk of injury is low, any injuries occurring during experimentation should be reported to the Research Team so that they can be dealt with in line with Cardiff University policies. There are no arrangements for compensation if you are harmed by taking part in this research but if you are harmed as a result of someone's negligence, then you may have grounds for legal action, but you may have to pay for it. Regardless of this, if you wish to make a complaint, or have any concerns about the way you have been approached or treated during the research you can lodge a complaint through the appropriate University channels.

10. What will happen to the results of the research study?

The results of this study will be presented amongst scientific and engineering peers at conferences. The data will remain anonymised at all times while in the public domain. If you are interested in the results of the experiments it is possible to contact the researchers with the contact details below.

11. Who is organising and funding the research?

The research is being organised by the Institute of Medical Engineering and Medical Physics (IMEMP) which is part of the Cardiff University School of Engineering. The research expenses for this project are being met by the Cardiff University School of Engineering.

12. Contact for Further Information

Mr Ramesh Swaminathan
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Cardiff University
Queen’s Building
The Parade
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CF24 3AA
Wales, UK.
Email: swaminathanr@cf.ac.uk

Appendix B:

Thesis Publications

Journal of Back and Musculoskeletal Rehabilitation -1 (2015) 1–7 1
DOI 10.3233/BMR-150626
IOS Press

The prediction of neck extensor force using surface electromyography

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[b] *Faculty of Health and Social Sciences, Bournemouth University, Bournemouth, Dorset, UK*

Abstract.
BACKGROUND: The relationship between muscular force and electromyography (EMG) has been investigated by numerous researchers. EMG has not previously been used as a means of estimating force in the cervical erector spinae (CES).
OBJECTIVE: Use EMG of the CES musculature to indirectly predict neck extension force.
METHODS: Isometric contractions of the CES muscles were studied at increasing levels of contractile force across all participants ($n = 12$) to produce an individualised force-EMG relationship. The method of least squares was used to determine the linear regression trend line for the force-EMG relationship. The validity of these individual 'correlation curves' was demonstrated through further, blinded, investigation.
RESULTS: A linear relationship was identified for the individualised correlation curves that gained in strength for $< 50\%$ maximum voluntary contraction (MVC; $R^2 > 0.8$ for 80% of trials). The prediction of muscle force from the correlation curves was found to be statistically similar to the equivalent experimental data ($p > 0.05$). Given the tendency of EMG to slightly overestimate force in most cases, an adjustment coefficient was calculated to reduce the error in the predicted force data.
CONCLUSIONS: This study reports a validated method using EMG to indirectly acquire CES muscular force, which has application for clinicians and research scientists working in fields including sport and rehabilitation.

Keywords: EMG, force, muscle, cervical spine

1. Introduction

The quantification of force has long been of interest to the clinician and sport scientist, providing information to determine both muscle health and the loading tolerance of joints [1] and soft tissues [2]. The measurement of force is commonplace in biomechanics laboratories, for example where force plates determine ground reaction force [3]; however force plates are rare in clinical and sporting environments. Particularly in sporting environments, force has a strong correlation with the failure of musculoskeletal structures [4–6]. Force measurement in these environments often takes the form of a manually held load cell applied to the limb to determine maximal force [7]. This method can easily be applied to many regions of the body. When investigating spinal biomechanics this method relies on the ability to place the load cell on the head or trunk. There are occasions when this method proves impossible, for example in attempting to measure cervical muscle force during lifting, or in more dynamic sporting encounters. In this scenario, the load cell would need to track the moving segment, thus introducing errors and experimental difficulties. Furthermore, there are occasions when the interest is in submaximal force, a common requirement for activities of daily living [8,9]. Owing to these limitations, an alternative method for acquiring this data is necessary.

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One viable option to estimate force may be to use EMG which measures the electrical activity of muscle. The relationship between EMG and force has been of interest to researchers for a number of decades [10], though there remains disagreement in whether such a relationship is linear [11–13]. If a relationship can be established between force and EMG, then it may prove possible to use EMG alone to predict muscle force in relatively controlled environments, an approach which has been suggested for the lumbar spine [14,15]. Such a method could then offer a solution to the measurement of force when traditional load cell or force plate methods are inappropriate thus, removing the aforementioned environmental and technical constraints. This would be useful for sports scientists, biomechanists and clinicians who encounter such environmental constraints.

This study describes a method to use EMG to predict muscle force in the cervical spine.

2. Methods

2.1. Subjects

Twelve male participants were recruited from Cardiff University. Exclusion criteria included a history of any spinal injury, or any indication of neuromusculoskeletal neck problems, such as limited range of motion or pain. The mean (SD) age of the participants was 25.8 years (3.59), mean (SD) height was 1.78 m (0.07), mean SD weight 77.9 kg (11.1) and mean (SD) BMI was 24.47 kgm^{-2} (2.33). This study was approved by the Cardiff School of Engineering Ethics Committee, with all volunteers providing written consent.

2.2. Instrumentation

A force platform (PS-2141, PASCO, California, USA) was secured in an inverted position, to measure force generation of the neck extensor muscles. EMG data was collected using a portable, wireless EMG system (PS850, Biometrics Ltd. UK) with a bandwidth of 20–450 Hz and common mode rejection ratio of 110 dB at 60 Hz. All raw signals collected by the system were pre-amplified with a gain of 1000, with data sampled at 1000 Hz. SX230 bipolar EMG electrodes (Biometrics Ltd. UK) were used, with a fixed inter-electrode distance of 20 mm. Electrodes were attached using double-sided adhesive tape (T350, Biometrics Ltd. UK) to unprepared skin, as per the manufacturer's

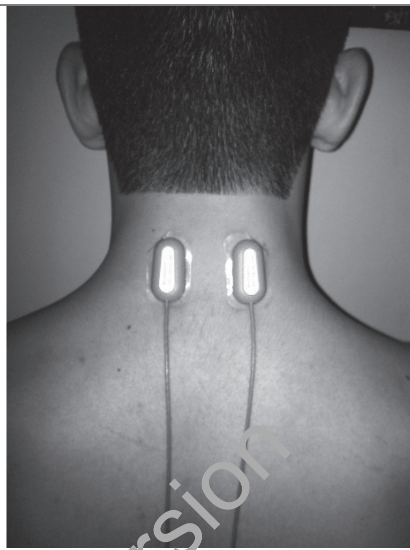

Fig. 1. Positions of bipolar electrodes on the left and right cervical erector spinae muscles.

recommendation. A reference electrode was attached over the ulnar styloid and two electrodes were placed 10 mm bilaterally at the C4/C5 level, between the anterior margin of the trapezius and the midline of the muscle body (i.e. on the left and right cervical erector spinae (CES) in line with the muscle fibres, Fig. 1) [9,16].

2.3. Experimental procedure

Participants were instructed to perform a series of isometric neck muscle contractions by pushing upwards with the back of their head on the force platform, with gradually increasing contraction. Participants adopted a position of 90° hip flexion, placing their hands on a bench for support when performing contractions, but were instructed not to use their arms to aid force production. When performing the contractions, participants were asked to fix their eyes on a point marked on the floor to minimise the amount of sagittal plane cervical motion. Furthermore, participants were instructed to perform the contractions as if they were attempting to tilt their head to look upwards in order to activate the muscles of interest. Firstly, participants were allowed to warm up by performing a number of motions in each direction. They then practised contractions of gradually increasing force, characterised as minimum, 25% maximum voluntary contraction (MVC), 50% MVC, 75% MVC and MVC (i.e. *incremental* contraction data). Exact representation of these contractions was not required, as it was a range

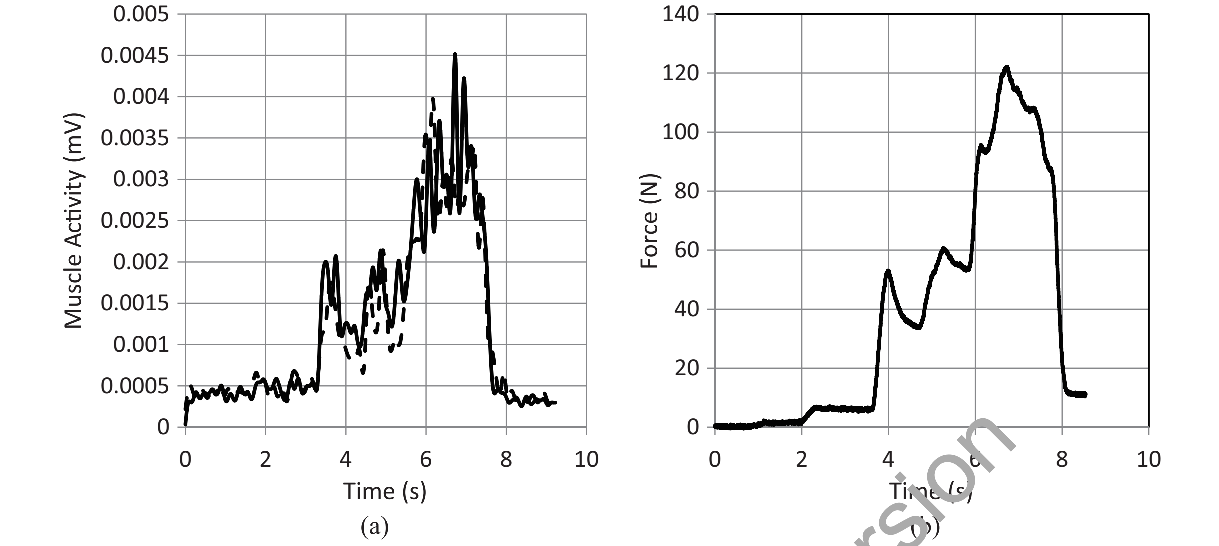

Fig. 2. Graphical display of (a) filtered right and left CES activity (solid line – left CES activity; dashed line – right CES activity) and (b) vertical force production of the neck extensor muscles for Participant 2, trial 1.

of contractions from minimum to maximum which was of interest. Each contraction was held momentarily, before increasing to the next contraction, with the overall trial repeated three times. A 120 second delay between each trial allowed for full physiological recovery [17]. This data served to explore the relationship between force and EMG and was used for the regression analysis. Following this, each participant performed self-randomised contractions (i.e. *randomised* contraction data), again repeating three trials and this data was used for the validation. The principle researcher was blinded to the force data for the *randomised* contraction trials.

2.4. Data processing

All data was processed in Matlab (Mathworks, 2012a). The raw EMG data was demeaned, full wave rectified and low pass filtered using a zero-lag 4^{th} order Butterworth filter with a cut-off frequency of 5 Hz, to produce a linear envelope for each channel. The raw *incremental* force data was magnitude normalised in Matlab to start at zero.

Peaks in the *incremental* muscle activity and muscle force data were identified and plotted. Given the linearity of the majority of such correlations [18,19], a linear regression trend line was plotted for both EMG channels (versus muscular force) using the method of least squares. Correlation coefficients (R^2) were calculated to determine the strength of the linear relationship for each participant.

2.5. Validation

The *randomised* EMG and force data was used to investigate the accuracy and, thus, validity of the protocol. The *randomised* EMG data was processed as above with the addition of time normalising both EMG and force data. Muscular force was predicted by reading off against the personalised correlation curves. The investigator remained blinded to the force data recorded. Comparison between the predicted and actual muscular forces was then performed, calculating the absolute and percentage errors for both left and right-sided EMG data. Left, right and an average of predicted muscle force were then plotted against actual force for each participant.

2.6. Statistical analysis

The Bland-Altman method [20] was used to plot the difference between predicted force and actual force for each participant, whilst the mean difference and upper and lower limits of agreement were also plotted. A two-tailed match paired t-test was used to explore for significant differences between predicted and actual force for all data points, and an analysis of the mean error was performed to determine the 95% confidence intervals (CI).

3. Results

An example of EMG and equivalent force data are presented in Figs 2 and 3 respectively to highlight iden-

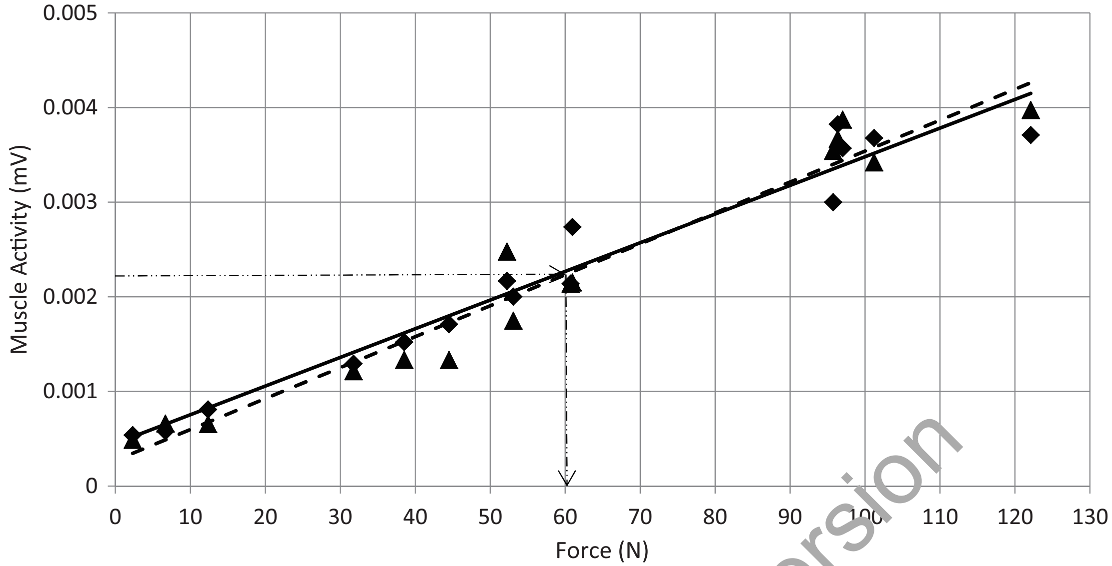

Fig. 3. Linear relationship of force and EMG for all 3 trials for a typical participant. Diamond – Left side; Square – Right side. Solid line – Left side trend line, $R^2 = 0.9544$; dotted line – Right side linear trend, $R^2 = 0.9556$.

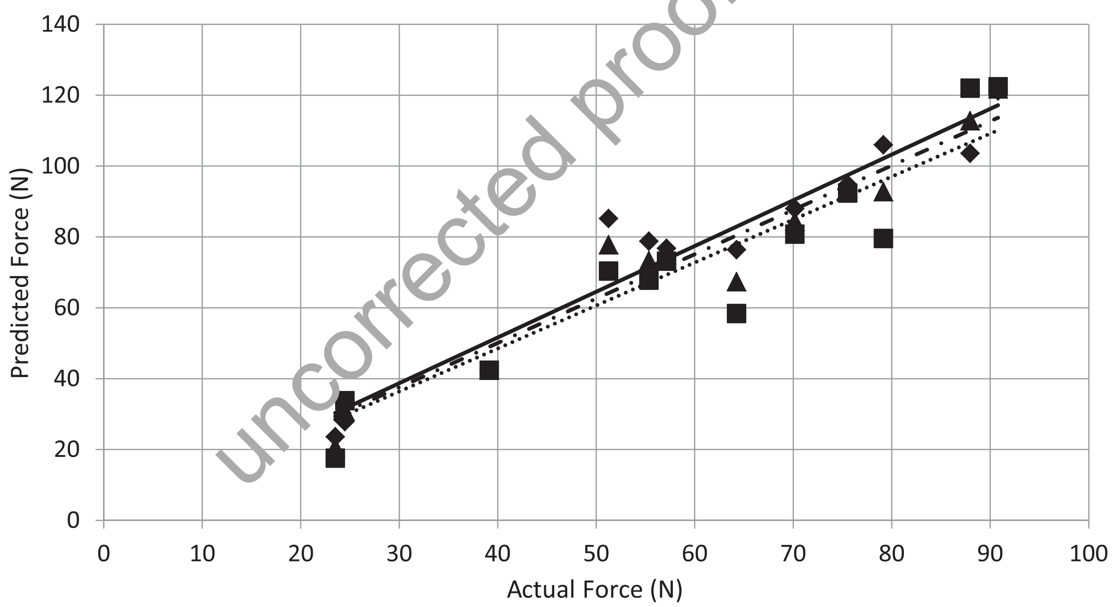

Fig. 4. Predicted force against actual force for left and right sides and average of both sides. Diamond – Left side; Square – Right side; Triangle – Average. Solid line – Left side trend line; dotted line – Right side linear trend line; Dash and dotted line – Average linear trend line. ($R^2 = 0.9296$ (left side); $R^2 = 0.8828$ (right side); $R^2 = 0.9394$ (average)).

tification of the 5 peaks of one particular trial. All 5 force peaks and maximum EMG activity were easily identifiable, as can be seen in the figures. Data displayed in Figs 2–4 are from the same participant.

A linear line of best fit was superimposed and individual force-EMG graphs were used to determine predicted force (Fig. 3). The correlation between force and EMG data ranged from $R^2 = 0.771$–0.954 for the left and $R^2 = 0.472$–0.967 for the right side. Combining the sides by averaging resulted in $R^2 = 0.734$–0.996. In excess of 80% of R^2 values exceeded 0.8, indicating a strong linear relationship between force and EMG. The data points for contractions less than 50% MVC had significantly less deviation from the linear trend

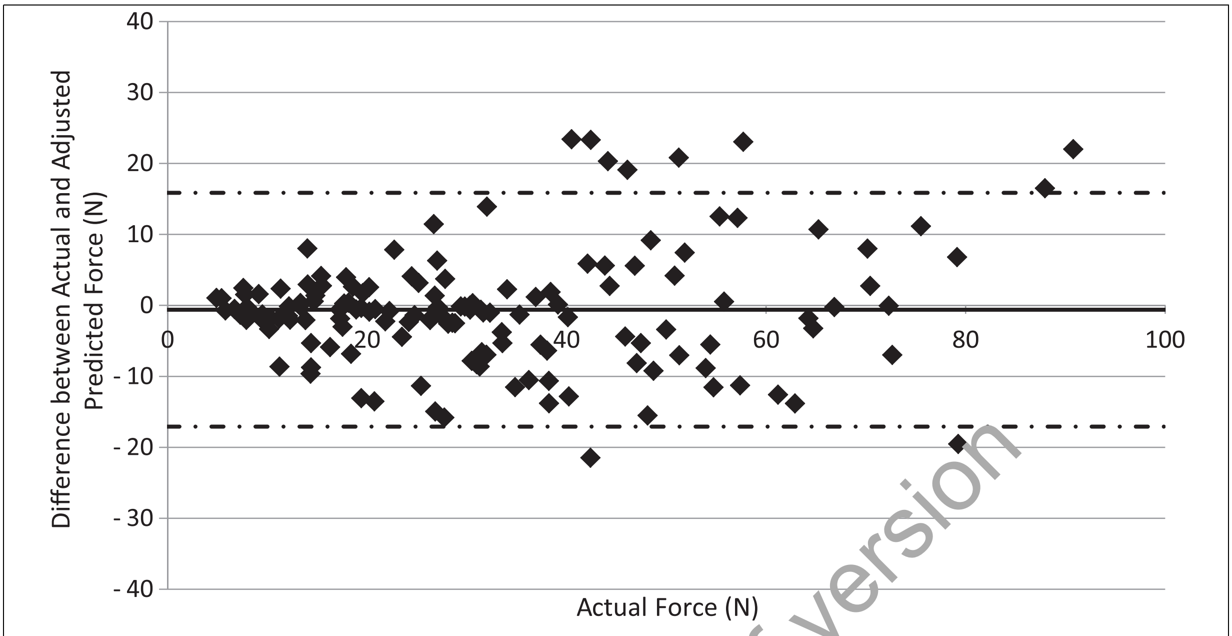

Fig. 5. Bland-Altman plot of differences (actual force vs difference in adjusted predicted force) for all 12 participants.

line than for contractions greater than 50% MVC. A polynomial model was also fitted to the data to determine which model would fit best to the data but the R^2 values for the polynomial were lower. Therefore, a linear model was chosen.

Predicted and actual force measurements for the *randomised* blinded trials are presented in Fig. 4. There were no significant differences between predicted and actual force for all participants ($t = 1.598$, $p = 0.112$). The predicted force magnitude provided an overestimate in most cases. Owing to the high level of consistency in this error (systematic error) the mean ratio between the predicted and actual force values was determined and used as a constant to adjust predicted values. The use of this mean constant resulted in improved estimation and a much-improved p-value ($t = -0.889$, $p = 0.375$). The mean absolute difference for all data points between the actual and adjusted predicted muscle force was 5.80 N (95% CI 4.84–6.76 N). Converting this to a percentage, the mean difference was 18.68% (95% CI 15.98–21.39%).

A Bland-Altman [20] plot was used to graphically represent the level of agreement between adjusted predicted and actual CES muscular force (Fig. 5), with the mean difference represented by the solid line. The dashed lines indicate the upper and lower limits of agreement – that is, two standard deviations either side of the mean.

4. Discussion

This study demonstrates the use of EMG to predict muscle force in the cervical region. The results demonstrate no significant difference between the actual force and that predicted by EMG. The high correlation coefficient reported in this study provides evidence of a linear relationship between CES force and EMG activation (Fig. 3), which is consistent with previous studies considering other muscles groups [11,12]. It is, however, in contradiction of other studies where a non-linear relationship was observed [13]. Both linear and non-linear models were fitted to the data and a linear model was found to provide the best correlations and therefore was the method chosen for the current study. The correlation coefficients were stronger than those describing polynomial trend, which was also fitted to these data during preliminary data analysis. The linearity of the correlation strengthened at forces < 35N (Fig. 4), which is consistent with the relationship described by previous researchers [18,19]. The strongest correlations were obtained for mean values of the right and left CES musculature; hence, the mean value is recommended when measuring EMG data for symmetrical tasks.

The results demonstrate a frequent over-prediction of force from EMG. This consistent direction of error enables a simple constant to be calculated and used to adjust the predicted values. The absolute mean errors between predicted and actual force were enhanced by the adjustment constant, therefore this adjustment is recommended for future application of such a method. The frequent over-prediction observed during the self-randomised contractions may be because these contractions had more dynamic properties than those used for the correlation trials. The mean absolute difference values were less than 6 N. This guides clinicians and sports scientists with understanding the degree of confidence with which EMG can be used to predict muscle force. The consistency of this protocol is evidenced by 90% of data points lying within the acceptable limit of error (Fig. 5).

The results of this study suggest that the method outlined above could be used with confidence when the aim is quantifying the muscle force. This is especially important in a rehabilitation setting where gradual loading of damaged structures should form part of the rehabilitation protocols [21–23]. Furthermore, it is known that paraspinal muscles alone can generate enough force to cause injury, including spinal fracture even in healthy spines [24–26]. Therefore, knowledge of muscle forces is important for rehabilitation, especially in regions of structural weakness following injury. The quantification of the effect of exercises and rehabilitation protocols on neck force will enable clinicians to make better choices as to appropriate exercises to be included and when. The method described above could be one way of further developing this knowledge and optimising the recovery of patients.

Individualised correlation curves were used in this study to predict muscular force production in recognition of the large variation in EMG amplitude that hinders the transferability of correlations between people [12]. This approach did, however, introduce inter-participant variation with, for example, differences in the range of contractions from different persons causing a clustering of data points at lower and higher ends of some force-EMG curves. Additionally, each participant was instructed to perform isometric contractions, to facilitate identification of a relationship. Adopting this technique does, however, prove difficult to isolate specific muscles for EMG measurements, meaning that the surrounding muscles may have been inadvertently recruited to support in the CES contraction, thus negatively affecting the EMG signal. Furthermore, the adoption of a position of 90° hip flexion may mean that back extensors are also used to aid force production. It is recognised that this methodology is for isometric and not dynamic contractions and thus will be more applicable to environments where there is very little motion of the neck. It may prove a useful protocol however for sporting environments where direct force measurement is impossible and therefore an indirect measurement system is required. For example, a potential application could be during rugby scrummaging where there is a relatively small amount of motion of the neck and players also adopt a position of 90° hip flexion. EMG signals have a number of factors that can affect it and, as such, it is acknowledged that EMG-force correlation trials will have to be performed prior to each testing session without removing or changing the position of the electrodes. Carrying out a correlation trial prior to performing the action in question would provide a correlation that is specific to the conditions of that particular day and the participant.

5. Conclusions

This study reports the development and validation of a method for measuring CES muscular force production, indirectly, using EMG. This protocol will prove useful for other investigators with an interest in spinal biomechanics, or those who are working in sports or rehabilitation, to quantify muscular force production of the CES muscles in non-laboratory situations where direct measurement of force production proves impossible.

Acknowledgments

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Conflicts of interest

None.

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A kinematic analysis of the spine during rugby scrummaging on natural and synthetic turfs

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A kinematic analysis of the spine during rugby scrummaging on natural and synthetic turfs

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ABSTRACT

Artificial surfaces are now an established alternative to grass (natural) surfaces in rugby union. Little is known, however, about their potential to reduce injury. This study characterises the spinal kinematics of rugby union hookers during scrummaging on third-generation synthetic (3G) and natural pitches. The spine was sectioned into five segments, with inertial sensors providing three-dimensional kinematic data sampled at 40 Hz/sensor. Twenty-two adult, male community club and university-level hookers were recruited. An equal number were analysed whilst scrummaging on natural or synthetic turf. Players scrummaging on synthetic turf demonstrated less angular velocity in the lower thoracic spine for right and left lateral bending and right rotation. The general reduction in the range of motion and velocities, extrapolated over a prolonged playing career, may mean that the synthetic turf could result in fewer degenerative injuries. It should be noted, however, that this conclusion considers only the scrummaging scenario.

ARTICLE HISTORY
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KEYWORDS
Rugby; scrum; spine; playing surface; kinematic

Introduction

There is an increasing rise in the popularity of rugby union worldwide. In parallel, World Rugby, formerly known as the International Rugby Board (IRB), continually reviews the laws to ensure player safety and spectator enjoyment. One area of the game that has come under particular scrutiny is the scrum. A drive to improve player safety and facilitate a quick, safe and fair restart following minor infringements (IRB, 2013) has resulted in significant rule changes. The complex interaction of the 16 players, combined with the desire to regain ball possession by being a superior scrummaging team, means that the scrum is a demanding physical environment. Consequently, whilst the scrum typically forms a relatively small period of game-time, this environment is associated with a disproportionality high percentage of injuries (6–13%) (Brooks, Fuller, Kemp, & Reddin, 2005; Fuller, Brooks, Cancea, Hall, & Kemp, 2007; Schick, Molloy, & Wiley, 2008; Taylor, Fuller, & Molloy, 2011). The lower leg and shoulder musculature are frequently injured in front-row players as a result of scrummaging and account for up to 54% and 66% of these two types of injury, respectively (Brooks, Fuller, & Kemp, 2005; Brooks & Kemp, 2011). Scrummaging also causes a significant proportion of all rugby spinal injuries (Bohu et al., 2009; Quarrie, Cantu, & Chalmers, 2002; Secin, Poggi, Luzuriaga, & Laffaye, 1999; Wetzler, Akpata, Laughlin, & Levy, 1998). Furthermore, the scrum is associated with a relatively high injury risk compared with other contact events (Fuller, Brooks, Cancea, et al., 2007; Taylor, Kemp, Trewartha, & Stokes, 2014) and collapsed scrums are associated with an even greater injury risk (Roberts, Trewartha, England, & Stokes, 2014). The most common and well-documented injuries are to the cervical spine, both chronic and acute (Bohu et al., 2009; Dunn & van der Spuy, 2010; Secin et al., 1999; Wetzler et al., 1998), but injuries to the lumbar region as a result of scrummaging have also been reported, which include lumbar disc injury or radiological abnormalities (Fuller, Brooks, & Kemp, 2007; Iwamoto, Abe, Tsukimura, & Wakano, 2005). Front-row players represent as high as 78% of all cervical spine injuries in the scrum (Bohu et al., 2009; Brooks, Fuller, Kemp, 2005) with the "hooker" position at the greatest statistical risk ($P < 0.01$) (Bohu et al., 2009; Secin et al., 1999; Wetzler, Akpata, Albert, Foster, & Levy, 1996; Wetzler et al., 1998).

Scrum stability is an integral part of player safety, as an unstable scrum may expose front-row forwards to scenarios that are potentially dangerous (Williams & McKibbin, 1987). Greater vertical and lateral forces may be generated (Milburn & O'Shea, 1994) and greater excursion of range of motion (ROM) means players must make more postural adjustments and are thus themselves less stable (Cazzola, Preatoni, Stokes, England, & Trewartha, 2015).

World Rugby has also focussed on improving game quality by permitting synthetic turf for use at all playing levels, ensuring a consistent playing surface standard and so encouraging high-quality and faster paced rugby. In the UK, elite teams including Cardiff Blues, London Saracens and Newcastle Falcons have adopted such surfaces now using a "3rd generation" (3G) synthetic turf that comprises a stone base, shock pad, carpet and rubber infill. Such surfaces are specifically designed to more accurately replicate the mechanical response of natural turf (IRB, 2003), thereby eradicating the extenuated ball bounce and high injury prevalence associated

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with earlier generations. The injury prevalence on synthetic turfs is perceived to be higher than on natural turf; however, no significant differences have been recorded in the literature (Ekstrand, Timpka, & Hägglund, 2006; Fuller, Clarke, & Molloy, 2010; Steffen, Andersen, & Bahr, 2007; Williams, Trewartha, Kemp, Michell, & Stokes, 2015). Ekstrand et al. (2006) reported only insignificant differences relating to ankle sprain incidence, whilst foot, ankle and knee injuries all had a greater, though statistically insignificant, prevalence during synthetic turf gameplay (Fuller et al., 2010). Indeed, no researcher has yet been able to identify a definitive cause and effect relationship relating to play on synthetic turf (Ekstrand et al., 2006; Fuller, Dick, Corlette, & Schmalz, 2007; Steffen et al., 2007).

A high prevalence of injury within the scrum has already prompted other studies to focus on quantifying force production during "machine-based" scrummaging (Milburn, 1990; Preatoni, Stokes, England, & Trewartha, 2013; Quarrie & Wilson, 2000) and, most recently, "live" scrummaging (Cazzola et al., 2015) environments. The presented study therefore addresses an important gap within the literature of a comprehensive analysis of player spinal kinematics during scrummaging on natural versus synthetic turf. This study aimed to quantify spinal kinematics during live scrummaging on two different playing surfaces with specific focus on the hooker as a result of the distribution of injuries throughout the scrum. Given that there is no significant change in injury incidence between the two types of turf, this study hypothesises that no significant changes will occur in spinal kinematics during scrummaging of the playing position investigated.

Methodology

Participants

Twenty-two participants were recruited from a convenience sample of local community club and university teams. The participants were divided equally into two groups, with the playing surface composition (i.e. natural or synthetic/3G turf) dictated by the surface available to a particular team. Teams only had access to either natural or synthetic turfs, and thus the study was unmatched as a result of the availability of the playing surface. All participants played in the hooker position and had been appropriately trained to play in the front row according to the discretion of the team's qualified coach (IRB, 2013). Exclusion criteria included those players with inadequate front-row playing experience according to the coach based on World Rugby guidelines (IRB, 2013), a history of any major spinal injury or any indication of current neuromusculoskeletal neck problems (e.g. pain). The World Rugby laws do not specifically state what inadequate experience is; this is at the discretion of the qualified coach. Recordings were undertaken of age, height, body mass, neck, shoulder and chest circumference (anthropometric data), number of training sessions per week, years of playing experience and number of scrummages per week (background data). No significant differences ($P > 0.05$) were found for any of the anthropometric or background information collected between the two groups. The study was approved by the Cardiff School of Engineering Ethics Committee, with all volunteers providing written consent.

Procedures

Data acquisition

Six landmarks were identified (forehead and the spinous processes of C7, T7, T12, L3 and S1) to create five spinal segments defined as the cervical (C), upper thoracic (UTx), lower thoracic (LTx), upper lumbar (ULx) and lower lumbar (LLx) regions. All landmarks were identified through palpation with the exception of the forehead, where a consistent position was identified across all players using a specially modified scrum cup. C7 was palpated by identifying the bony prominences of C6 and C7 and getting the participants to extend their neck. By doing this, C6 glides away and C7 remains prominent (Middleditch & Oliver, 2005). T7 was found by identifying the inferior borders of the scapulae and finding a midpoint of a line drawn in the transverse plane connecting these points (Willems, Jull, & Ng, 1996). T12 was identified by counting up from L4. L4 was identified by a line bisecting in the transverse plane at the most superior point of the iliac crests (Burton, 1986). L3 was found in a similar manner by counting up from L4. S1 was found by finding the midpoint of a line in the transverse plane created by the posterior superior iliac spines (Chakraverty, Pynsent, & Isaacs, 2007). Three-dimensional kinematic data describing these five spinal segments were measured using a string of six inertial sensors (ThetaMetrix, Waterlooville, UK). The sensors sampled at 40 Hz/sensor, with data recorded via USB to a laptop computer for retrospective analysis. The sensors were attached to the skin using hypoallergenic double-sided tape, with all trailing wires secured using Hypafix tape (BSN Medical, Hamburg, Germany).

Experimental procedure

All trials were conducted outdoors and were part of their team's training session. All participants completed their club's warm-up routine, before being instrumented with the six inertial sensors. All participants first took part in a series of ROM trials to quantify their normal, active spinal ROM. This was performed in a predefined order for the full spine, the cervical spine in a standing position and the cervical spine in a position of hip flexion similar to that of scrummaging. During each stage of the ROM trials, each movement was repeated three times in the order of flexion, extension, right and left lateral bending and finally right and left rotation. Between each movement, the participant resumed a neutral position. For each of these motions, the peak ROM was calculated for each motion and each segment and collated for the two groups.

Having performed the ROM trials, the participant was then joined by another seven players to comprise a complete "pack". An opposing pack was drawn from other suitably experienced players from within the same club. Each scrum was performed using the current engagement sequence of "crouch-bind-set" (CBS), dictated by the trainer leading the session. This scrum engagement sequence was introduced worldwide in the 2013–2014 playing season. On the "crouch" call, the front row must bend at the hips and be ready to engage. On the "bind" call, the props must take a grip of their opponent's jersey. On the "set" call, the scrums are permitted to engage through the interlocking of the heads of the front row (IRB, 2013). Three live scrums were performed, with players being given adequate rest between each trial. Data recording commenced when the players adopted the scrum position and were ready to engage with the opposing pack. Players were instructed to scrummage as per typical "live" training sessions, and to wear appropriate attire (including boots, shorts and a shirt). After each trial the sensors were checked to ensure proper adhesion to the skin and, if needed, realigned to counteract any problems with movement. Any trial where the sensors had moved or become detached was deleted.

Having collected the data of absolute orientation, described as Euler angles, this was converted into rotation matrices and the resultant angles between two adjacent sensors were calculated through matrix multiplication to determine the motion of each individual spinal segment, through a custom written code in Matlab (Lee, Laprade, & Fung, 2003; Williams, Haq, & Lee, 2013a). The rotation order corresponded to rotation describing flexion/extension, lateral bending and then rotation. This yielded six different motions for each segment, which were defined as flexion, extension, left lateral bending, right lateral bending, left rotation and right rotation. Peak ROM values were extracted over the full duration of the scrum. Specific time periods of the scrum were not identified. ROM data was filtered using a low-pass, bidirectional Butterworth filter with a cut-off frequency of 6 Hz to remove high-frequency noise (Fioretti, 1996). The neutral position was defined as the standing position adopted at the start of the ROM trials, serving as the reference plane for all subsequent data; hence, a position of 30° upper lumbar flexion is 30° relative to the standing position. A five-point differentiation method was used to yield the angular velocity of each individual segment for the six different motions.

Statistical analysis

Mean and standard deviations of peak kinematic variables were calculated for each condition (synthetic vs. natural turf) and these pooled data were used to determine the existence of any significant differences. An analysis of the data was performed (SPSS 18, SPSS Inc., Chicago, USA) to test for normality. A Pearson's correlation test was used to determine whether any correlation existed between peak ROM of the groups and the anthropometric and background data collected. A two-tailed, independent *t*-test (with Bonferroni correction) was performed across every motion of every segment, considering the ROM and angular velocity ($P < 0.05$). Furthermore, Cohen's effect size (d) was calculated to determine the magnitude of differences between conditions and $d > 0.8$ was considered to be a "large" effect.

Results

Anthropometry

Table 1 shows the participant's height, mass and BMI.

No significant difference was determined between the groups for the anthropometric or background data ($P > 0.05$).

Table 1. Anthropometric data for both natural (n = 11) and synthetic turf (n = 11) groups.

	Natural turf group	Synthetic turf group
Age (years)	24.73 (4.49)	22.08 (3.78)
Height (m)	1.78 (0.04)	1.76 (0.05)
Mass (kg)	99.63 (8.57)	98.00 (13.37)
BMI (kg · m^{-2})	31.52 (2.65)	31.51 (4.38)

Note: Mean data are presented with standard deviation in parentheses.

ROM

The mean ROM data for the natural and synthetic turf groups are presented in Table 2. No significant differences were identified across the peak motion of any segment between the two groups. Furthermore, there was no correlation between the peak ROM, anthropometric data and playing history.

Hooker spinal kinematics during live scrummaging

Mean peak ROM and angular velocity magnitudes were calculated for each group and all 30 movements (i.e. six motions for each of the five segments). Figure 1 displays mean, peak cervical ROM for all six motions on natural (dark grey shading) and synthetic (diagonal lines) turfs, respectively. This data indicates a reduction in right rotation, when comparing the mean peak ROM when scrummaging on natural versus synthetic turfs. This, however, was not statistically significant ($P > 0.01$), but the size of the effect was moderate ($d > 0.5$).

Table 3 displays mean peak percentage ROM relative to maximum mean peak ROM for all 11 participants of each group on natural and synthetic turf, respectively. Peak percentage ROM was calculated from the mean peak ROM experienced during scrummaging for the group and the mean peak ROM of the group demonstrated during the ROM trials. No significant differences were present in the ROM between the two groups ($P > 0.05$) for any segment during the live scrummaging trials.

Regardless of surface, it is apparent that scrummaging utilises almost all the available ROM in the upper lumbar spine for flexion. Similarly, almost all available right rotation of the upper lumbar segment was utilised and a large amount of both right lateral bending and right rotation of the lower lumbar segment. These percentages, however, are for the peak values of ROM and thus these segments are only periodical in such a position.

Figure 2 shows an example of the dynamic time history of motion for the upper lumbar segment during live scrummaging (CBS) on (a) synthetic and (b) natural surfaces. Owing to the dynamic and unpredictable nature of scrummaging, the peaks extracted from the ROM graphs for the analysis did not always appear at the same time points in the data. This particular segment was always in a position of flexion throughout the trials.

For angular velocity, the lower thoracic segment demonstrated a reduction from scrums on natural turf to scrums on synthetic turf. These differences were in left and right lateral bending and left rotation, but none of these were significant after Bonferroni correction. When considering the size of the effect, however, Cohen's d-value was above 0.8 for all three of

Table 2. Mean peak ROM for the natural (n = 11) and synthetic (n = 11) turf trials.

	Flexion (°)		Extension (°)		Right lateral bending (°)		Left lateral bending (°)		Right rotation (°)		Left rotation (°)	
	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural
Cervical (Scrummaging position)	36.2 (6.6)	42.4 (7.4)	23.6 (10.0)	27.7 (8.6)	28.1 (10.1)	30.4 (9.5)	27.3 (8.2)	30.4 (8.3)	42.2 (10.0)	45.2 (6.5)	45.1 (11.6)	43.4 (5.8)
Cervical	46.1 (9.8)	43.2 (7.2)	33.6 (9.4)	33.0 (7.8)	35.5 (4.6)	34.9 (4.3)	30.2 (4.9)	34.4 (7.0)	42.1 (11.3)	42.9 (6.1)	41.9 (8.2)	44.9 (7.1)
Upper thoracic	17.7 (12.7)	23.7 (10.4)	10.8 (11.9)	1.5 (1.6)	9.4 (3.1)	10.4 (4.0)	8.1 (4.6)	9.6 (4.2)	26.9 (12.9)	23.3 (11.01)	20.5 (9.4)	24.8 (10.2)
Lower thoracic	16.3 (9.1)	19.27 (7.4)	7.3 (6.0)	6.3 (5.3)	15.5 (9.1)	15.8 (3.2)	16.1 (7.6)	16.0 (2.7)	27.8 (14.4)	33.3 (7.9)	28.9 (14.2)	31.0 (8.2)
Upper lumbar	34.1 (11.9)	33.4 (7.9)	12.9 (5.9)	13.6 (7.5)	13.9 (4.2)	14.5 (4.3)	12.3 (4.4)	15.1 (7.4)	9.4 (5.9)	12.9 (8.8)	10.7 (6.6)	13.3 (7.6)
Lower lumbar	24.4 (13.2)	23.4 (10.1)	20.1 (13.9)	22.3 (9.6)	8.3 (4.5)	9.2 (4.4)	7.1 (2.6)	9.8 (5.3)	9.7 (5.3)	9.2 (2.7)	7.3 (3.3)	11.9 (3.2)

Note: Mean data (degrees) are presented with standard deviation in parentheses.

the aforementioned variables, indicating that the size of the effect was large.

Discussion

The high injury prevalence of the rugby scrum has ensured this fundamental component of the game – and specifically force generation – has long remained a research focus (Cazzola et al., 2015; Milburn, 1987; Quarrie & Wilson, 2000). The current study reflects the sport's evolutionary nature, and is the first to compare the hooker's spinal kinematics in scrums on natural versus synthetic turf.

Hooker peak ROM during the scrum did not change significantly with respect to the playing surface; however, the synthetic surface produced a more conservative ROM in nearly 80% of the 30 variables. The natural surface produced greater angular velocities in right lateral bending and left rotation, and left lateral bending, in the lower thoracic segment. Although these differences were not statistically significant after Bonferroni correction, the size of the effect was large ($d > 0.8$).

Whilst no injury data was collected during this study, the results suggest that there may be a trend towards a slightly reduced injury risk when scrummaging on synthetic surfaces due to greater stability. Spinal angular velocity has a direct relationship to trunk muscle activity with increased angular velocities resulting in much increased paraspinal muscle activity (Fan, Liu, & Ni, 2014; Mawston & Boocock, 2012; Williams, Haq, & Lee, 2013b). Increased paraspinal muscle activity has been shown to cause spinal compression in both cervical (Skrzypiec, Pollintine, Przybyla, Dolan, & Adams, 2007) and lumbar (Adams & Hutton, 1982, 1985a) regions. Similarly, in the thoracic spine, increased paraspinal muscle activity has been suggested to cause greater compressive loading (Caneiro et al., 2010). Compressive loading of the thoracic spine significantly loads the cortical shell with 45% of the load being borne by this structure (Kilincer et al., 2007). In the lumbar region, compressive loading has been shown to be at a high risk of endplate fracture and, when combined with bending, may cause injury to the intervertebral disc (Adams & Hutton, 1982, 1985a). Furthermore, there is a relatively high prevalence of thoracic spine injuries reported in rugby forwards from T8-T12 (Hind, Birrell, & Beck, 2014). In the cervical spine, compression causes the loss of intervertebral disc height and resultant increased load bearing on the neural arch and uncovertebral joints (Skrzypiec et al., 2007). Over a prolonged time, this may lead to the development of degenerative changes in the spine such as the formation of osteophytes (Kumaresan, Yoganandan, Pintar, Maiman, & Goel, 2001). These degenerative changes have been observed previously in front-row players in the cervical spine (Berge, Marque, Vital, Senegas, & Caille, 1999; Scher, 1990). Whilst it is not known whether similar pathologies occur in the thoracic spine, it appears likely that owing to the responses shown to similar increases in muscle activity of other regions of the spine, similarity can be predicted for the effects on the thoracic spine.

As this was one of the first studies to investigate spinal kinematics during live scrummaging, the measured ROM

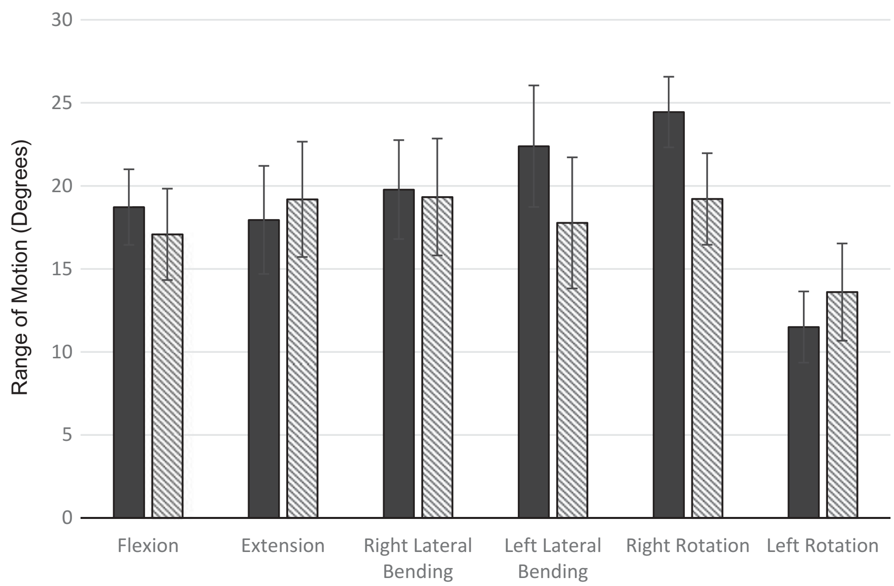

Figure 1. Mean peak cervical ROM during live scrummaging (CBS engagement) on natural (dark grey shading) and synthetic (diagonal lines) turf for all 11 players of each group.

Table 3. Mean peak percentage ROM during live scrummaging (CBS) for the synthetic (n = 11) and natural (n = 11) turf trials.

	Flexion		Extension		Right lateral bending		Left lateral bending		Right rotation		Left rotation	
	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural
Cervical	47.2	44.5	81.4	64.8	68.7	65.0	65.2	73.6	45.5	54.1	30.2	26.5
Upper thoracic	46.4	37.1	40.4	46.0	79.7	77.1	70.9	77.3	40.7	62.2	94.6	71.6
Lower thoracic	50.7	23.8	65.3	36.4	32.1	47.3	74.6	47.5	38.6	39.9	29.5	42.4
Upper lumbar	88.4	98.0	1.3	3.8	58.9	66.2	55.6	72.3	97.2	95.7	68.1	67.2
Lower lumbar	42.6	78.5	56.5	42.7	80.8	91.9	75.2	93.6	79.1	20.7	57.1	72.9

merits further discussion. Stand-alone ROM values can be difficult to interpret and hence these values were converted to a percentage of maximum ROM. It is well documented that the strength of the spine is compromised at the end of the range; therefore, loading the spine at, or close to, the end of range significantly increases the risk of spinal failure. The results illustrate a very high percentage of total range was used during scrummaging for specific segments. Over 95% and 90% of the total range of upper lumbar flexion and rotation was observed, respectively, a position shown to result in significant motion segment weakness (Gallagher, Marras, Litsky, & Burr, 2006), be associated with reduced paraspinal muscle activity and a transfer of loads from active to passive tissues (McGill & Kippers, 1994). The lower lumbar spine also appears to utilise a significant amount of its available range, ~85% of lateral bending, a position known to compromise the pars interarticularis (Stokes, 1988). Therefore, it appears that scrummaging may place the spine in a position that results in its compromised osteoligamentous strength and may be one factor associated with the high spinal injury prevalence relating to scrummaging.

Although there is evidence to suggest that loading the spine towards the end of its range has a significant risk of injury, as mentioned above, there is also evidence that suggests that more constrained kinematic conditions, as seen on synthetic turf, may lead to repeated stresses on the same vertebral structures (Adams & Hutton, 1985b; Adams, McNally, Chinn, & Dolan, 1994; Adams, McNally, & Dolan, 1996). Therefore, the link between more constrained kinematic conditions (i.e. synthetic turf) and reduced injury risk is not quite as straightforward as it may initially seem. For example, flexion reduces stress in the apophyseal joints and posterior half of the annulus fibrosus, but it also increases stress on the anterior annulus (Adams & Hutton, 1985b). Thus, the suggestion that scrummaging on a synthetic surface may reduce injury risk must be interpreted with caution owing to the aforementioned reasons.

The cervical spine is widely reported in the literature to suffer from the greatest number of injuries during scrummaging (Quarrie et al., 2002; Scher, 1982; Wetzler et al., 1998) and front-row players are well documented to suffer from premature chronic degeneration of the cervical vertebrae (Berge et al., 1999; Broughton, 1993; O'Brien, 1996; Scher, 1990). The data presented relating to angular velocities (Table 4) of the cervical spine is relevant to this as it may provide some information as to why these chronic injuries are so prevalent. From the data, it can be seen that mean peak cervical spine angular velocity was greater on natural turf than on synthetic turf. For flexion, left lateral bending and right rotation, there was a medium effect size (d > 0.5). Greater angular velocities

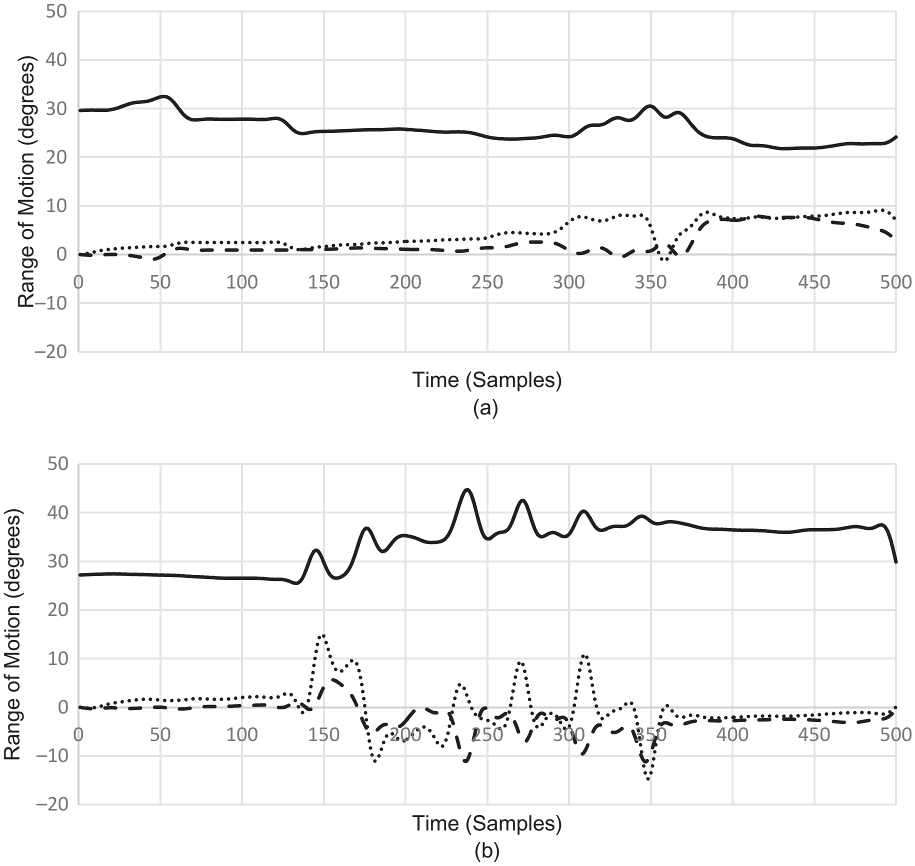

Figure 2. Example of dynamic upper lumbar ROM during live scrummaging (CBS) on (a) synthetic and (b) natural turfs. Solid line – flexion-extension; dashed line – lateral bending; dotted line – rotation.

Table 4. Mean peak angular velocity during live scrummaging (CBS) on 3G and natural surfaces.

	Flexion ($° \cdot s^{-1}$)		Extension ($° \cdot s^{-1}$)		Right lateral bending ($° \cdot s^{-1}$)		Left lateral bending ($° \cdot s^{-1}$)		Right rotation ($° \cdot s^{-1}$)		Left rotation ($° \cdot s^{-1}$)	
	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural	Synthetic	Natural
Cervical	8.5 (4.3)	10.3 (4.2)	9.4 (4.2)	10.9 (6.7)	8.2 (3.7)	14.9 (13.0)	7.6 (2.0)	7.1 (1.9)	7.1 (1.9)	10.5 (6.4)	8.2 (4.0)	9.2 (5.1)
Upper thoracic	6.6 (2.2)	10.1 (6.9)	6.9 (3.7)	6.1 (1.4)	6.0 (1.3)	8.0 (4.7)	6.7 (3.8)	9.4 (5.0)	7.6 (5.1)	10.6 (9.7)	7.6 (2.1)	12.0 (12.8)
Lower thoracic	3.0 (1.6)	4.4 (2.3)	2.8 (1.8)	4.5 (2.8)	***3.8 (2.6)***	***7.1 (3.5)***	***3.3 (1.8)***	***6.0 (1.9)***	4.8 (3.5)	7.5 (4.9)	***4.4 (2.1)***	***9.8 (5.9)***
Upper lumbar	4.1 (2.3)	5.4 (2.7)	3.0 (1.6)	4.6 (2.4)	4.1 (1.8)	5.0 (2.2)	3.4 (2.1)	5.9 (4.4)	4.2 (3.2)	6.0 (5.3)	5.0 (4.6)	7.8 (4.3)
Lower lumbar	4.3 (3.5)	5.7 (6.5)	4.2 (2.4)	9.2 (8.7)	3.9 (2.7)	6.0 (5.4)	3.7 (1.9)	5.7 (4.1)	4.8 (4.6)	8.7 (5.5)	3.9 (2.2)	9.2 (7.8)

Note: Mean peak data ($° \cdot s^{-1}$) are presented with standard deviations in parentheses. Large effect sizes (d > 0.8) are highlighted in bold italics.

mean greater loading/force of the structure in question (Yoganandan & Pintar, 1997; Yoganandan, Pintar, Cusick, & Hollowell, 1999; Yoganandan, Pintar, Sances Jr, Reinartz, & Larson, 1991). The repetitive loading experienced during scrummaging (Scher, 1990), with forces, (Nightingale, McElhaney, Camacho, Winkelstein, & Myers, 1997; Nightingale, Richardson, & Myers, 1997; Yoganandan et al., 1991) velocities and accelerations (Portero, Quaine, Cahouet, Thoumie, & Portero, 2013; Yoganandan et al., 1991) that are comparable to the current data and other published data on scrummaging (Cazzola et al., 2015; Milburn, 1993; Preatoni et al., 2013; Preatoni, Stokes, England, & Trewartha, 2015; Quarrie & Wilson, 2000) may, with time, lead to chronic degenerative changes and neck pain (Berge et al., 1999; Lark & McCarthy, 2010; Pinsault, Anxionnaz, & Vuillerme, 2010; Scher, 1990). Thus, a reduction in angular velocity is likely to be a positive outcome for the playing position considered as it may delay the onset of chronic degenerative changes that are often seen in these players.

Scrum stability is extremely important to try and reduce the number of collapses and therefore reduce the risk of catastrophic spinal injury. Stability was estimated by considering the magnitude of kinematic variables where lower magnitudes were taken to mean more stability for the player being investigated. This approach is similar to that adopted by Cazzola et al. (2015), where lower excursions/ROM were considered to mean greater stability, since players made less postural adjustments. There has been some anecdotal evidence to suggest that scrums are more stable on synthetic surfaces, as there was an observed decrease in the number of collapsed scrums (BBC, 2013), but this is the first study to provide empirical evidence to suggest that scrummaging on a synthetic surface does have a potentially positive effect on stability for the player that was investigated.

Strengths and limitations

This was the first study to investigate spinal kinematics of multiple segments during rugby scrummaging on different turfs. To the best of the authors ability, as many variables as possible were controlled to make the statistical tests as robust as possible. Variables that were controlled included the hooker always being on the attacking side, the same coach was used to call the engagement sequence and the same scrum-half was used.

Ideally, the groups in this study would not have been unmatched. It would have been preferable to have a group of players who had access to training on both synthetic and natural turf pitches so that each player was exposed to both conditions, meaning a more robust statistical test could have been used. Owing to the availability of pitches to different teams, it was not possible to do this and this limitation is acknowledged by the authors.

Conclusions

A study was conducted to assess the differences in spinal kinematics of the rugby union hooker when scrummaging on two different playing surfaces: synthetic and natural. A general trend was observed of a reduction in the magnitude of kinematic variables for hookers in both ROM and angular velocity, but no significant differences were observed. Only lower thoracic angular velocity, of left and right lateral bending and left rotation proved to have a large effect size ($d > 0.8$). These reductions suggest that there is more stability when scrummaging on synthetic pitches compared to natural turf, as players have lower excursions and therefore fewer postural adjustments. Given the fact that research suggests there to be no difference in traction on synthetic and natural surfaces, the authors suggest that it is the consistency in the properties of synthetic turf in a variety of weather conditions that give rise to this increased spinal stability of the hooker. No significant differences ($P > 0.05$) were found between normal ROM values and anthropometric and background data, between the two groups, and no correlation between ROM and anthropometric and background data were observed. Although separate groups were used, the lack of significant differences of anthropometric and background data observed between the two suggest similarity between the groups. Finally, the observed reduction in peak kinematic variable magnitudes suggests that scrummaging on synthetic pitches may potentially be safer, in the long term, as a result of the increased stability that this data suggests.

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Disclosure statement

The authors would like to declare that there is no conflict of interest.

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Does the new rugby union scrum sequence positively influence the hooker's in situ spinal kinematics?

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ABSTRACT

Background: Scrummaging is unique to rugby union and involves 2 'packs' of 8 players competing to regain ball possession. Intending to serve as a quick and safe method to restart the game, injury prevalence during scrummaging necessitates further evaluation of this environment.

Aims: The aim of this study was to determine the effect of scrummage engagement sequences on spinal kinematics of the hooker. The conditions investigated were: (1) live competitive scrummaging using the new 'crouch, bind, set' sequence; (2) live competitive scrummaging using the old 'crouch touch pause engage' sequence and (3) training scrummaging using a scrum machine.

Methods: Inertial sensors provided three-dimensional kinematic data across 5 spinal regions. Participants (n=29) were adult, male community club and university-level hookers.

Results: Engagement sequence had no effect on resultant kinematics of any spinal region. Machine scrummaging resulted in lesser magnitudes of motion in the upper spinal regions. Around two-thirds of the total available cervical motion was utilised during live scrummaging.

Conclusions: This study indicates that the most recent laws do not influence the spinal kinematics of the hooker during live scrummaging; however, there may be other benefits from these law changes that fall outside the scope of this investigation.

What are the new findings?

- There are no spinal kinematic differences between the two competitive scrummaging scenarios indicating neither a positive nor negative effect of the law change.
- Large magnitudes of thoracic and lumbar spine motion may put these spinal regions at an increased risk of injury.
- Machine scrummaging is a much more constrained kinematic environment.
- Modest amount of cervical flexion may be owing to a large amount of 'stabilising' muscle forces in order to minimise head displacement.

INTRODUCTION

Rugby union players are consistently exposed to a relatively high risk of injury across all playing levels.[1–12] While the high level of participation suggests that players accept this risk of injury, specific elements within the game remain the target for reduction strategies. The scrummage (hereafter termed the 'scrum'), involves eight players from each team attempting, en masse, to push their opponents backwards and regain ball possession. This represents the contact event with the highest risk of injury within the sport.[13] World Rugby, the game's governing body, has recently evolved the laws in an attempt to reduce this risk of injury; hence, as of the 2013/2014 playing season, the scrum followed the referee's command 'crouch-bind-set' (CBS), as opposed to 'crouch-touch-pause-engage' (CTPE).[14]

Scrum injuries can be either acute, chronic or degenerative in nature,[15–17] with longer term exposure to this demanding biomechanical environment associated with disc narrowing (35–71% incidence), osteophyte formation (83%), apophyseal joint degeneration (74%), and degeneration of the vertebral endplates (77%).[18 19] Given the time spent scrummaging compared with other contact events during a game, it is associated with a disproportionately high percentage (6–13%) of all spinal injuries within the sport.[20–25] These spinal injuries predominantly happen to players of the front row. Front-row players (ie, the three players from each team, who directly oppose each other) are at particular risk of both chronic and acute injuries, accounting for 78% of all scrum-based acute cervical spine injuries.[21 26] While it is acknowledged that risk of injury is multifactorial, the biomechanical demands of the scrum are unique and are likely to significantly contribute to the risk of injury. The scrum consists primarily of the 'hit/impact', followed by a sustained attempt to 'shove'. The 'hit' is where the two packs initially engage, immediately after the referee's instruction. Both packs then produce a sustained shove, aiming to push the opposition away from the ball. As the packs engage, the shoulders of the front-row players of each team collide, their heads become interlocked and are forced underneath the chest of their opposing player. The impact force of this interaction has previously been reported, varying from 4.4–16.5 kN,[27–31] measured by instrumenting either the scrum machine or shoulder pads (point of impact) of front-row players, during live scrummaging.[31]

Recent kinematic studies have additionally focused on investigating neuromuscular activation patterns during simulated, live and machine-based scrummaging.[31 32] While no significant differences were observed between engagement sequences for body kinematics, it does appear that the 'crouch-bind-set' engagement sequence may prepare the cervical spine by stiffening the joints prior to impact. Furthermore, it was reported that machine scrummaging does not accurately replicate live scrummaging, suggesting the need for future studies to focus on investigations of live scrummaging.

As the scrum has been identified as being specifically associated with injury, a greater understanding of spinal kinematics will further aid in the quest for injury reduction. To date, while studies have considered machine-based and simulated live scrummaging,[33 34] technical challenges have prevented studies from acquiring spinal kinematics during competitive scrummaging. Given the association between scrummaging and the potential for spinal injury, it is critical that these challenges are met to enable data acquisition in the most relevant physical environment.

This study aimed to determine the resultant spinal kinematics of the hooker during *competitive* scrummaging using two different sequences (CBS and CTPE), and using CBS when performing machine-based scrummaging. This study focused on the most injury-prone player, the hooker,[21 23–25] generating the first data sets that allow direct comparison of spinal kinematics between the new and old scrum sequences. It was hypothesised that the hooker's spinal kinematics will be more constrained during the new (CBS), as opposed to the older (CTPE), engagement sequence.

MATERIALS AND METHODS

A repeated-measures design was employed to evaluate the effect of the scrum engagement sequence (within-group factor) on three-dimensional spinal kinematics. The methodology replicated the scrum law evolution in both training and competitive scenarios.

Participants

Hookers were recruited from a convenience sample of local community club and university rugby union teams. Data were collected from all participants during team versus team interactions (hereafter described as competitive scrummaging), using both the old and new scrum sequences. Some of these hookers were also analysed when their eight players (the 'pack') were using a scrum machine within a typical training scenario. All participants were deemed by their qualified coach to have the requisite skill and knowledge of playing in this position, and had been appropriately trained, according to World Rugby guidelines.[14] Potential participants were excluded if they had inadequate front-row experience, a history of any major spinal injury, or any indication of current neuromusculoskeletal neck problems (eg, pain). The study was approved by the Cardiff School of Engineering Ethics Committee, with all volunteers providing written informed consent.

Equipment

Participants were instrumented with a string of inertial sensors (THETAMetrix, Waterlooville, UK) adhered to the skin on the forehead and overlying the spinous processes of C7, T7, T12, L3 and S1 (figure 1). Each sensor was comprised of a triaxial accelerometer, gyroscope and magnetometer, and sampled at 40 Hz/sensor. The

Figure 1 Inertial sensor placements over the spinous processes of C7, T7, T12, L3 and S1.

sensors had previously been validated for reliability and accuracy during factory calibration, with errors typically <1°. The sensors' fusion algorithm, incorporating Kalman filtering, provided drift-free computation of absolute orientation. Similar sensors have been widely used for spinal motion analysis,[35 36] and such an experimental set has been used previously during rugby scrummaging.[37] Each sensor was adhered directly to the skin using double-sided hypoallergenic tape.

This created five spinal 'segments' (ie, the spinal region between these 6 specific locations), which were defined as the cervical (C), upper thoracic (UTx), lower thoracic (LTx), upper lumbar (ULx) and lower lumbar (LLx) regions. A modified scrum cap was used to reinforce the attachment of the forehead sensor. Absolute orientations, expressed as Euler angles, were determined for each sensor and stored on a PC.

Procedures

Data collection was integrated within scheduled training sessions, such that all players were familiar with the facilities and environment. All participants completed their club's standard warm-up routine before being instrumented with the inertial sensors and taking part in their trial. Each trial lasted approximately 15 min. Prior to scrummaging, each player completed range of motion (ROM) trials to characterise their normal, active, spinal ROM. This was performed for the full spine, while additional cervical spine ROM data was collected in a position of hip flexion (similar to that of scrummaging). Each uniplanar motion was performed three times in a predefined order (flexion/extension, lateral bending and rotation), with a short pause in between each motion. The participant always resumed a neutral position between movements. For the cervical segment, normalisation was performed from the hip flexion ROM trial. For all other segments, normalisation was performed from the standing ROM trial. The peak ROM was calculated for each motion and each spinal segment. All participants then performed three trials of the new (CBS) and old (CTPE) sequences, within a competitive environment, with scrums being performed on either grass or synthetic (3G) turf, depending on their training facilities. The sequence of scrums was randomised, with players instructed to replicate their 'in-game' scrum performance. Prior to testing, participants were assigned a value, via a table of randomised numbers, as to the order they would perform the two engagement sequences. All trials for one engagement sequence were completed before moving onto the next sequence. Players were allowed 2 min for full physiological recovery between each trial,[38 39] to minimise the effect of fatigue.[40 41] Additionally, where a scrum machine was available, the pack also performed the CBS sequence to replicate a typical training scenario.

Data processing

Absolute orientations of the sensors, described as Euler angles, were converted to rotation matrices, and resultant angles between two adjacent sensors were calculated through matrix multiplication. This process is commonplace in three-dimensional kinematic analysis, with the resultant angles describing the ROM of each spinal segment. ROM computation was completed using custom Matlab (Mathworks, 2012a) scripts.[35 37 42] The rotation order corresponded to motion describing flexion-extension, lateral bending and rotation. This resulted in six motions for each spinal segment described as flexion, extension, left and right lateral bending, and left and right rotation.

ROM values were used to determine the spinal ROM relative to each participant's maximal ROM, defined in the earlier ROM trials, as it is increasingly acknowledged that end-range postures have the potential to increase risk of injury.[43–46] Data were time-normalised according to event duration. All kinematic data were filtered using a zero lag 4th order low-pass Butterworth filter, with a 6 Hz cut-off frequency.[47] Sensor-skin adhesion was confirmed before and after each trial, with data discounted in cases where there appeared to have been sensor displacement.

Participants and environment

Table 1 describes the 29 hookers analysed during the competitive scrum scenarios. Fourteen of these hookers were also analysed within a training scenario, with their pack scrummaging against a machine on a grass surface.

Statistical analysis

Normality and sphericity were checked using Shapiro-Wilk and Mauchly's test, respectively. A one-way repeated measures ANOVA, with scrum condition as the within-group factor (ie, competitive (CBS), competitive (CTPE), training), was applied to test for differences across the variables, with post-hoc Bonferroni comparisons where appropriate. Significance was set at $p<0.05$. All tests were performed with SPSS V.22 (SPSS Inc, Chicago, USA).

RESULTS

Kinematic analysis

There were no significant differences in kinematics for any spinal region between the two engagement

Table 1 Data describing the 29 hookers analysed within a competitive scrum

Descriptor	Mean (SD)
Age	23.4 (4.2) years
Height	1.76 (0.04) m
Mass	101.1 (12.8) kg
Body mass index	32.6 (4.0) kg/m^2

Table 2 Mean (SD) data describing the ROM for each sequence

Spinal region	ROM (degrees)	Competitive (n=29) CTPE	CBS	Training (n=14)
Cervical	Flexion	16.6 (10.6)	18.1 (9.3)	11.1 (8.2)
	Extension	16.9 (11.2)	17.8 (9.9)	18.2 (5.5)
	Right side flexion	20.8 (11.3)	18.5 (9.8)	16.1 (8.7)
	Left side flexion*	19.1 (13.3)*	19.5 (12.3)*	9.9 (5.3)*
	Right rotation*	24.9 (10.0)*	20.9 (9.1)	14.3 (9.7)*
	Left rotation	14.8 (10.6)	12.8 (7.8)	17.4 (9.1)
Upper thoracic	Flexion	10.6 (5.4)	8.9 (7.3)	9.0 (5.4)
	Extension	25.1 (13.5)	22.1 (10.0)	13.6 (5.3)
	Right side flexion	11.4 (6.8)	10.6 (6.2)	8.3 (5.0)
	Left side flexion	15.8 (9.9)	15.3 (11.8)	9.2 (6.1)
	Right rotation	10.7 (7.3)	14.7 (12.5)	10.6 (5.7)
	Left rotation*	21.7 (10.8)*	16.0 (9.6)	12.0 (11.1)*
Lower thoracic	Flexion	5.1 (5.4)	4.7 (6.6)	4.0 (4.9)
	Extension	17.4 (10.3)	16.0 (9.0)	16.1 (6.3)
	Right side flexion	7.6 (4.7)	6.7 (4.0)	8.5 (6.4)
	Left side flexion*	16.6 (5.5)*	15.3 (7.4)*	8.0 (3.8)*
	Right rotation	9.9 (7.2)	14.0 (11.6)	6.8 (3.7)
	Left rotation	13.2 (6.8)	12.0 (5.8)	9.4 (6.5)
Upper lumbar	Flexion	43.1 (12.3)	42.3 (11.0)	42.4 (10.3)
	Extension	0.8 (3.0)	0.4 (0.8)	0.8 (1.6)
	Right side flexion	7.9 (6.0)	7.2 (4.8)	6.9 (6.1)
	Left side flexion	10.9 (6.3)	10.6 (6.3)	7.6 (4.3)
	Right rotation	13.5 (5.9)	12.9 (7.9)	9.7 (9.9)
	Left rotation	7.7 (7.4)	8.0 (6.8)	6.4 (4.6)
Lower lumbar	Flexion	14.2 (12.5)	16.1 (17.1)	11.3 (12.0)
	Extension	9.8 (7.4)	11.6 (9.1)	9.9 (5.6)
	Right side flexion	7.5 (5.1)	9.4 (6.7)	7.2 (6.3)
	Left side flexion	8.3 (6.4)	10.1 (9.0)	7.9 (6.7)
	Right rotation	12.6 (10.1)	10.0 (10.9)	10.1 (8.6)
	Left rotation	11.9 (10.3)	15.9 (12.0)	10.0 (8.7)

*Statistically significant difference ($p<0.05$).
CBS, Crouch-bind-set; CTPE, Crouch-touch-pause-engage; ROM, range of motion.

sequences during competitive scrummaging ($p>0.05$). Table 2 presents the mean peak motion data for the two scrum engagement sequences.

Significantly greater ROM was identified in the upper three spinal segments in both competitive sequences, versus the training (machine scrummaging) scenario. No significant differences (ie, $p>0.05$) were evident in the two lowest spinal regions.

Data presented in figures 2A–C provides dynamic cervical ROM across the competitive CBS scrum scenarios, describing the flexion-extension, lateral bending and rotation left to right, respectively. Figures 2D–F describe equivalent data from the older CTPE engagement sequence. In each case, data has been time normalised and combined to yield a confidence corridor for ROM during scrummaging. The solid line represents the cohort mean and the dotted lines represent the upper and lower 95% CIs.

ROM relative to maximal ROM (figure 3) demonstrated no significant difference between scrummaging sequences ($p>0.05$). It did, however, identify that almost all available ROM was used for a number of spinal regions.

DISCUSSION

This study set out to determine whether the evolution of rugby laws relating to the scrum had any effect on hooker spinal kinematics during competitive scrums. The results presented here indicate that the changes in laws do not have any significant effect on hooker spinal kinematics when comparing between the CBS (new) and CTPE (old) scrummaging sequences. There were some kinematic differences between training and competitive scrummaging trials but only for the upper spinal regions. This study performed the first multiregional spinal analysis, generating data that enhances our understanding of spinal kinematics within both a training and competitive environment, and using the new and old scrummaging sequences.

Previous studies have demonstrated that CBS engagement has significantly influenced the biomechanics of scrummaging.[31 48] Indeed, growing evidence exists that the impact of engagement is reduced using CBS, as measured by the impact to the shoulder girdle in both machine,[49] and live[48 50] scrummaging. This was attributed to a reduced distance between front-row players prior to the impact phase. The current study did not

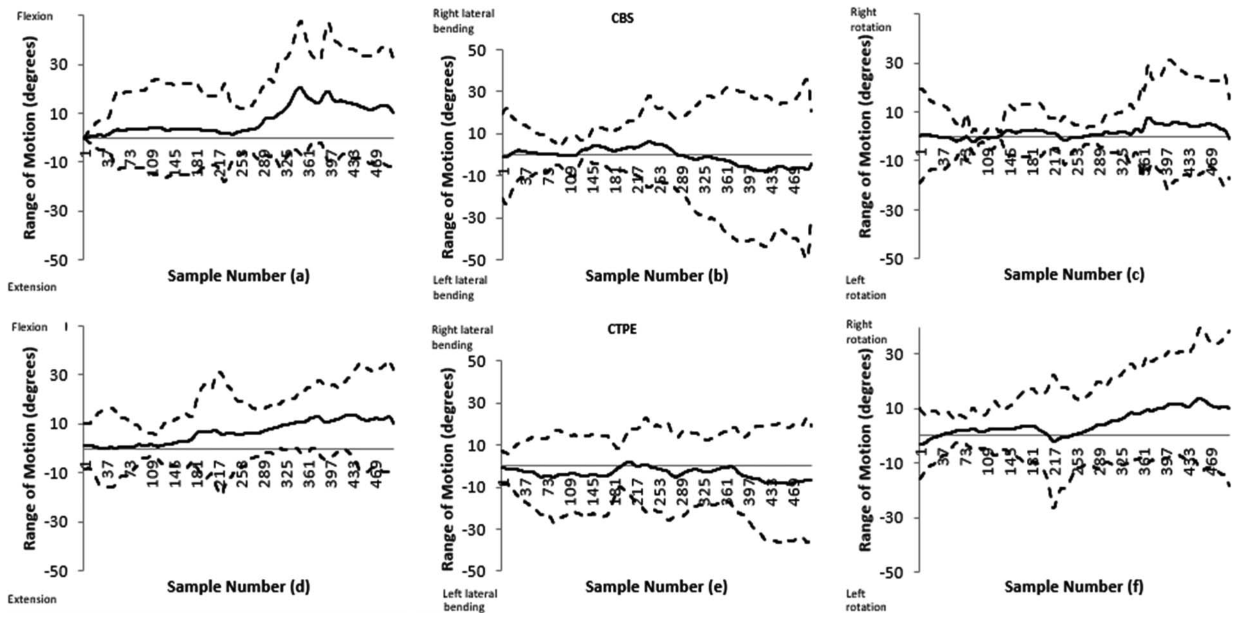

Figure 2 Dynamic cervical spine ROM during competitive (CBS and CTPE) training scrums. Solid line=mean; dashed lines= upper and lower 95% CI limits. CBS figures—(a: flexion/extension, b: lateral bending, c: rotation); CTPE figures—(d: flexion/ extension, e: lateral bending, f: rotation). CBS, Crouch-bind-set; CTPE, Crouch-touch-pause-engage; ROM, range of motion.

measure the impact phase specifically, nor impact at the shoulders. Instead, we sought further understanding of the law change on other biomechanical measures. To this end, the findings of the current study do not support or refute the recommendations to move to the new laws, though synthesising all available data would suggest that the new laws result in reduced shoulder impact, and do not negatively influence spinal kinematics.

The data presented in this study demonstrated no difference in spinal ROM across the scrummaging sequences. The extent of spinal motion varied across the population, and appeared to be dependent on both the participant and the nature of the scrum. The

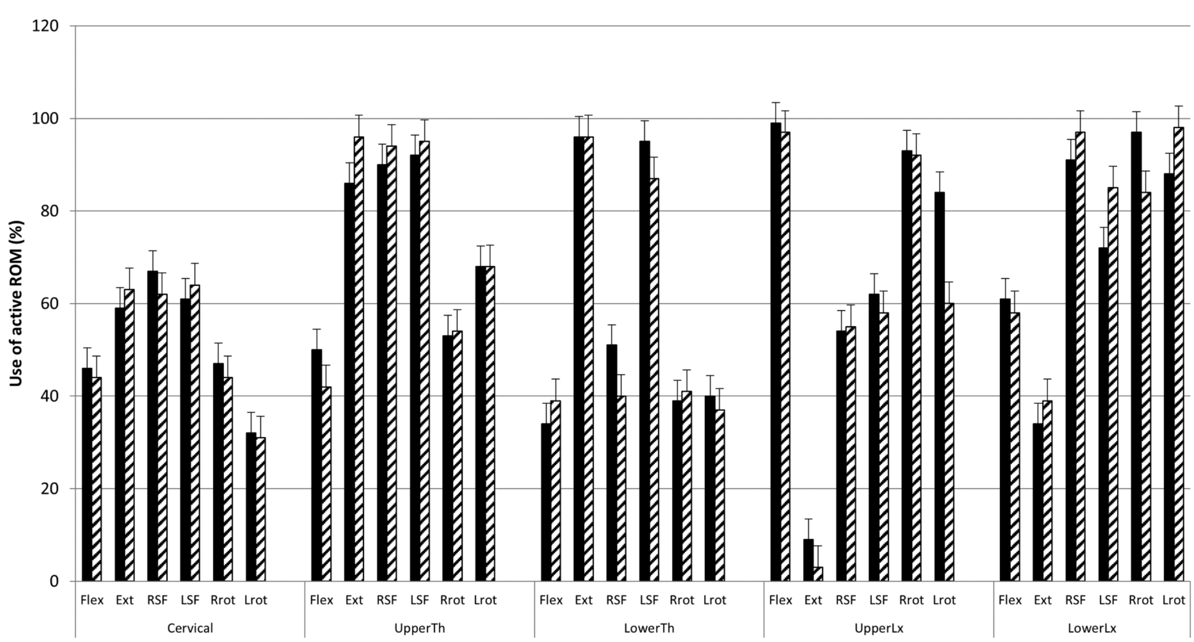

Figure 3 The relative proportion of active ROM used during competitive scrummaging per spinal region, averaged across the cohort. Black columns = CTPE; Hatched columns = CBS. Error bars represent SE. CBS, Crouch-bind-set; CTPE, Crouch-touch-pause-engage; ROM, range of motion.

particularly large variability observed in the cervical ROM (figure 2) would indicate that intrinsic control of ROM is the main determinant of kinematics during scrummaging; hence, it could be extrapolated that spinal kinematics may not be readily influenced solely by evolving scrum laws. It is also noted that approximately 60% of total sagittal ROM was utilised during scrummaging, suggesting that both sequences (CBS and CTPE) represent a relatively low risk of hyperflexion for most participants, a mechanism believed to be linked to catastrophic injury.[51] It may be that catastrophic hyperflexion injuries are the sole domain of the collapsed scrum; however, a more substantive claim can only be levied on observing a greater number of scrums.

Our data also indicated that machine scrummaging utilised significantly less ROM for the upper spinal regions, which is consistent with reports of less cervical muscle demand of machine versus live scrummaging.[32] Combining this data seems to suggest that machine scrummaging requires significantly less cervical and upper thoracic ROM, and less cervical muscle activity; hence, this appears to be an environment which poorly reflects the true cervical demands of live scrummaging. Indeed, knowledge of this less-intense environment may prove a useful addition to the rehabilitation pathway following cervical injury, while also highlighting the importance of additional conditioning prior to competitive scrummaging.

Reducing chronic injury from the scrum environment is complex. Greater or excessive spinal motion is associated with chronic degeneration in all regions of the spine,[52 53] thereby leading to conclusions regarding reduced ROM being beneficial for chronic injury potential. Indeed, we observed only a modest range of cervical motion (63% of total available range), suggesting that excessive motion may not be the main source of degenerative change in the cervical spines of front-row players. For a front-row player to constrain motion in an oppressing environment requires significant muscle activity. Such muscle activity would, of course, result in a cost of compressive load, also known to be linked to chronic degeneration.[54] The relatively modest cervical ROM observed in this study may be the result of a large amount of 'stabilising' muscle forces, where the demands of the dynamic competitive scrum see the hooker attempt to minimise head displacement. To this end, the greater muscle activity noted in previous CBS studies[32] would indicate relatively high compressive loads on the cervical spine.[54] This may provide some explanation as to the relationship between scrummaging and long-term degenerative changes in the cervical spine.

Scrummaging utilises a greater proportion of thoracic and lumbar spine ROM, versus cervical spine ROM. Indeed, over 90% of the available range was used during extension and lateral bending in the thoracic spine, flexion for the upper lumbar spine, and side bending and rotation for the lower lumbar spine. This is significant, as end-range spinal positions may result in reduced muscle activity,[55] altered muscle function,[56] increased load on the passive osteoligamentous spine,[57] and increased risk of tissue damage.[46] Indeed, reduced tissue compliance and bony opposition associated with end-range, negatively influences the available motion in other planes;[58 59] hence, the use of such large proportions of the available range may expose the thoracic and lumbar spine to greater risk of injury.[60]

In performing the first multiregional spinal kinematics evaluation of the hooker within the rugby union scrum, this study is unique in presenting the greatest detail of spinal kinematic data. This analysis revealed that, when considering the relative motion of the five spinal regions, there was statistically insignificant kinematic variation between the new (CBS) and old (CTPE) sequences. This study adds to the debate surrounding scrummaging, by using novel methods to report the first spinal kinematics from within a competitive environment. Our data does not, however, support the notion that the new scrum laws will succeed in reducing chronic spinal injury.

Limitations of the current study include that it was conducted during training, which is unlikely to represent an identical biomechanical loading pattern to the more aggressive and physical nature of a competitive match. No detail regarding the specific phases of scrummaging were captured; therefore, it was not possible to relate specific ROM to specific scrum phases. Owing to the sampling frequency of data capture, it was not possible to report on the impact forces experienced at the spinal regions, although this was not the primary aim of our study.

CONCLUSIONS

Our data indicates that the scrum engagement sequence does not affect spinal kinematics of the hooker during competitive scrummaging. Machine scrummaging results in less ROM for the upper spinal regions compared to competitive scrummaging. Approximately 60% of available flexion ROM of the cervical spine was used during competitive scrummaging, whereas the lumbar spine utilised the entire ROM. Our data adds to the debate that influencing spinal kinematics within the scrum may require more drastic changes in law, owing to an individual's intrinsic control of motion.

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