

**Canine lumbosacral fracture-luxation stabilised  
with the String of Pearls interlocking plate  
system or pins with polymethylmethacrylate:  
A biomechanical comparison**

by

**Johannes J. Nel**

Submitted in partial fulfilment of the requirements for the degree of

**MMedVet (Chir)(Small Animal Surgery)**

Department of Companion Animal Studies

Faculty of Veterinary Science

University of Pretoria

2016-01-20

**Canine lumbosacral fracture-luxation stabilised with the String of Pearls interlocking plate system or pins with polymethylmethacrylate:  
A biomechanical comparison**

**By:** Dr J.J.Nel  
Fourways Veterinary Hospital & Specialist referral centre  
c/o Witkoppen & The Straight  
Bryanston

**Supervisor:** Prof G.L. Coetzee  
Head of Section Small Animal Surgery  
Department Companion Animal Clinical Studies  
Faculty of Veterinary Science  
University of Pretoria

**Co-supervisor:** Dr C. Kat  
Department of Mechanical & Aeronautical  
Engineering  
University of Pretoria

**Co-worker:** Dr M. Hornsveld  
Department of Animal and Wildlife Services  
Faculty of Natural and Agricultural Science  
University of Pretoria

**Co-worker:** Dr P.J. van Staden  
Department of Statistics  
University of Pretoria

## DEDICATED TO:

- Professor Louis Coetzee, an excellent role model – both as a surgeon and as a person – for any aspiring student.
- Dr Neels du Plessis, a friend and mentor who patiently helped shape my surgical skills over the last few years.
- Ruan and Tiaan, my two sons, who stood by my side and sacrificed a lot of time which I could have spent with them during my years of studying.
- Leigh-Anne Liebenberg, my loving and caring girlfriend who patiently put up with me during my best and worst, and was always there to help and support me. Thank you for always believing in me, for the hugs of encouragement, and for the many litres of coffee you made for me during my studies.
- My supporting parents, who always believed in me and showed, by example, that through hard work and dedication, anything is possible.
- Most of all, our Heavenly Father, for the health and the blessings I receive every day which enable me to follow my dreams.



I, Johannes Jacobus Nel, hereby declare that the work on which this dissertation is based is original (except where acknowledgements indicate otherwise) and has not been previously submitted by me for another degree at this or any other University, Tertiary Education Institution, or examining Body.

2016-01-20

## ACKNOWLEDGEMENTS

Without the help of the following people, this dissertation would not have been possible:

- Prof. J.P. Schoeman – head of the Department of Companion Animal Clinical Studies, Faculty of Veterinary Science, University of Pretoria; for giving me the opportunity to complete my study in small animal surgery.
- Prof. G.L. Coetzee – my promoter and a brilliant surgeon who, through his passion for surgery, introduced many students to the joy of surgery. Thank you for the encouragement and the patient advice during my clinical studies.
- Dr Neels du Plessis – my veterinary promoter, for his valuable guidance in all the clinical aspects of my study.
- Dr Cor-Jacques Kat – my engineering co-promoter, for the many hours spent designing and building the test configuration, performing all the biomechanical testing, and collecting the data; and for his patience in explaining all the results to me.
- Dr Marius Hornsveld – for all his help with the anatomy of the lumbosacral area and his words of encouragement when I needed them most.
- Dr Paul van Staden and Mrs Joyce Jordaan – who performed the statistical analysis with enthusiasm and precision, making sense of the hundreds of sets of data collected.



- Sr. Leigh-Anne Liebenberg and Sr. Anria Wolmarans – for their help and patience in taking numerous radiographs for me.
- Fourways Veterinary Hospital and Specialist referral centre– for the financial support and the precious time they allowed me to take off in order to complete my studies.
- University of Pretoria – for the financial support needed to complete this project.
- National Council of SPCAs (NSPCA) – for approving my use of the cadavers for research purposes.

## TABLE OF CONTENTS

ACKNOWLEDGEMENTS .....	5
TABLE OF CONTENTS .....	7
SUMMARY .....	11
LIST OF TABLES.....	12
LIST OF FIGURES .....	13
LIST OF ABBREVIATIONS.....	16
CHAPTER 1: INTRODUCTION.....	18
1.1 Background .....	18
1.2 Problem statement .....	20
1.3 Aim.....	21
1.4 Hypothesis .....	21
1.5 Objective and value of this study .....	22
1.6 Conflict of interest.....	23
CHAPTER 2: LITERATURE REVIEW .....	24
2.1 Anatomy and Biomechanical considerations .....	24
2.2 Spinal fractures .....	25
2.3 Treatment of Fractures .....	27
2.4 Methods for stabilising vertebral fracture-luxations .....	29
2.4.1 Pin and PMMA.....	30
2.4.2 Vertebral body plating.....	33
2.4.3 SOP locking Plates.....	34



CHAPTER 3: MATERIALS AND METHODS.....	43
3.1 Introduction .....	43
3.2 Test Objective .....	44
3.3 Test methodology.....	45
3.4 Specimens .....	45
3.4.1 Inclusion criteria.....	45
3.4.2 Initial specimen preparation .....	46
3.4.3 Identification of the specimens.....	47
3.4.4 Storage of specimens .....	47
3.4.5 Anatomical components of the specimens.....	48
3.4.6 Assignment of the specimens .....	49
3.5 Radiographs and Photographs .....	49
3.6 Defect condition (Osteotomy of L7) .....	51
3.7 Fixation techniques (Osteotomy fixation).....	52
3.7.1 pin-PMMA.....	52
3.7.2 SOP Plating.....	57
3.8 Testing .....	61
3.9 Test equipment.....	62
3.9.1 Spinal loading simulator.....	63
3.9.2 Motion measurement system.....	64
3.10 Test conditions .....	64
<i>Temperature</i> .....	65
<i>Test duration</i> .....	65





<i>Moisture condition</i> .....	65
<i>Loading rate</i> .....	65
3.11 Specimen preparation .....	65
3.12 Preparation of the testing equipment.....	69
3.13 Loading of each bony specimen .....	70
3.14 Testing procedure .....	71
3.15 Data and statistical analysis .....	73
<u>3.15.1</u> Biomechanical parameters .....	73
<u>3.15.2</u> Statistical techniques used .....	77
<u>3.15.3</u> Data excluded from analysis and statistics.....	77
CHAPTER 4: RESULTS .....	78
4.1 Stability of the different fixation techniques.....	78
4.2 Strength of the fixation techniques.....	81
4.3 Modes of failure (MoF) .....	87
4.3.1 MoF1: Failure of joint L5-L6.....	87
4.3.2 MoF2: Failure of joint L6-L7 .....	90
4.3.3 MoF3: Failure (fracture) of the sacrum at the sacrum-epoxy interface .....	92
4.3.4 MoF4: Failure of the connection between the epoxy and the sacrum .....	95
CHAPTER 5: DISCUSSION.....	98
5.1 Introduction .....	98
5.2 Stability of fixation techniques .....	98
5.3 Strength of the fixation technique .....	99
5.4 Limitations of the study.....	103



CHAPTER 6: CONCLUSION .....	106
CHAPTER 7: RECOMMENDATIONS .....	107
REFERENCES .....	108
Appendix 1: Protocol Approval by the Research Committee. ....	115
Appendix 2: Protocol Approval by the Animal Ethics Committee. ....	116
Appendix 3: Approval from National Council of SPCA.....	117
Appendix 4: Copyright permission.....	117

## SUMMARY

A biomechanical comparison was conducted on two internal spinal fixation techniques: pins and Polymethylmethacrylate, and the String of Pearls Interlocking Plate System. Both techniques were applied to a surgically simulated level L7-S1 complete spinal injury, with the objective of the study being to compare the stability of the two techniques. Cadaver specimens from 18 skeletally mature large breed dogs were used. These specimens were randomly divided into two equal groups and fixated using one of the two internal spinal fixation techniques.

The lumbosacral spine specimens (L5-S3) were subjected to a mechanically applied bending moment, which was applied to the caudal and cranial ends of the specimen. Biomechanical parameters including range of motion (RoM), neutral zone (NZ), and elastic zone stiffness (EZS) were used to compare the stability of the two fixation techniques. No significant difference between the means of the NZ in flexion ( $p$ -value=0.3458), extension ( $p$ -value=0.1255), and the total value ( $p$ -value = 0.3458) of the injured lumbosacral (L7-S1) joint fixated with the two fixation techniques was found. Similarly, no significant difference between the means of the RoM in flexion ( $p$ -value = 0.2386) and extension ( $p$ -value = 0.1255), or between the means of the EZS in extension ( $p$ -value = 0.4094) was noted. Therefore, it can be concluded that the stability of the injured joint between the two fixation techniques is similar.

**Keywords:** lumbosacral joint; spinal fixation; biomechanical comparison; implant stability



## LIST OF TABLES

<b>Table 3.1</b>	Identification of specimens
<b>Table 3.2</b>	Assignment of specimens to the two fixation methods
<b>Table 3.3</b>	Load ranges used for testing fixated spine specimens
<b>Table 3.4</b>	Specimens and/or joint data which have been excluded from the analysis
<b>Table 4.1</b>	Biomechanical parameters of the three joints in the fixated spine specimens (Non-Parametric tests)
<b>Table 4.2</b>	Load range at which the PMMA fixated spines failed
<b>Table 4.3</b>	Load range at which the SOP fixated spines failed

## LIST OF FIGURES

- Figure 2.1** Lateral radiograph illustrating a typical lumbosacral fracture–luxation
- Figure 2.2** Photograph of a String of Pearls (SOP) Interlocking Plate (no screws)
- Figure 2.3** Close up view of a String of Pearls (SOP) Interlocking Plate
- Figure 2.4** SOP plate with bending tees inserted into some of the holes
- Figure 2.5** Caudo-lateral schematic view of a lumbar vertebra (L6), illustrating placement of the SOP plate and correct screw angle. (*Source: L. Liebenberg; Fourways Veterinary Hospital, Bryanston, South Africa*)
- Figure 2.6** Cranial view of L7 showing placement of recommended screw or pin Placement angle (cranial end of the vertebra is cut away to show the screws) (*Source: L. Liebenberg; Fourways Veterinary Hospital, Bryanston, South Africa*)
- Figure 2.7** Dorsal view of L6, L7, and sacrum depicting the specific entry points for correct screw or pin placement
- Figure 3.1** Dorsal view of a typical test specimen (L5-S3), before potting its cranial and caudal ends in epoxy
- Figure 3.2** Individual storage of the specimens
- Figure 3.3** Dorso-ventral radiographic view of the spinal segment L5 – S3
- Figure 3.4** Lateral radiographic view of the spinal segment L5 – S3
- Figure 3.5** Lateral radiograph illustrating a typical lumbosacral fracture–luxation
- Figure 3.6** Photo of the lateral view of the positive end-threaded pin placement.

- Figure 3.7** Photo of a right lateral view of a bone specimen after placement of the pins for the pin-PMMA method
- Figure 3.8** Photo of the dorsal view of the pin-PMMA placement
- Figure 3.9** Photo of the lateral view of the pin-PMMA placement
- Figure 3.10** Lateral radiographic view of the pin-PMMA placement
- Figure 3.11** Dorso-ventral radiographic view of the pin-PMMA placement
- Figure 3.12** Caudo-lateral schematic view of lumbar vertebra (L6), illustrating placement of the SOP plate and correct screw angle
- Figure 3.13** Photo of the dorsal view of the SOP placement
- Figure 3.14** Photo of the lateral view of the SOP placement
- Figure 3.15** Lateral radiographic view of the SOP placement
- Figure 3.16** Dorso-ventral radiographic view of the SOP placement
- Figure 3.17** Experimental setup used during the biomechanical testing of the spine specimens
- Figure 3.18** The Free Bending Canine Spine Spinal Loading Simulator (FBC-SLS)
- Figure 3.19** Potting and alignment of the sacral end of a bone specimen
- Figure 3.20** Potting and alignment of the cranial end of specimen
- Figure 3.21** Alignment of the top and bottom jaws of the FBC-SLS
- Figure 3.22** Position at which the weight of the BMT's and spine segment is measured
- Figure 3.23** Graph of cycles of the angle and bending moment of each joint with the associated ensemble average and standard deviation

- Figure 3.24** Graph depicting the bending moment- angle (Ens. Avg. (cycles)) characteristic of the three joints
- Figure 3.25** Graph depicting the load-displacement curves along which all specimens were evaluated
- Figure 4.1** Bending moment – angle characteristics of the PMMA and SOP fixated spines
- Figure 4.2** Displacement and bending moment vs. time of the pin-PMMA fixated specimen 13
- Figure 4.3** Displacement and bending moment vs. time of the pin-PMMA fixated specimen 1
- Figure 4.4** Photo of failure of joint L5-L6 (pin-PMMA)
- Figure 4.5** Photo of failure of joint L5-L6 (SOP)
- Figure 4.6** Graph of failure of L5-L6 joint (MoF1)
- Figure 4.7** Photo of failure of joint L6-L7 in pin-PMMA specimen
- Figure 4.8** Graph of failure of L6-L7 joint (MoF2)
- Figure 4.9** Photo of failure (fracture) of sacrum at sacrum-epoxy interface (pin-PMMA)
- Figure 4.10** Photo of failure (fracture) of sacrum at the sacrum- epoxy interface (SOP)
- Figure 4.11** Graph of failure (fracture) at the sacrum-epoxy interface (MoF3)
- Figure 4.12** Photo of failure of connection between epoxy and sacrum
- Figure 4.13** Graphs of failure of connection between epoxy and sacrum (MoF4)

## LIST OF ABBREVIATIONS

(+) Positive	-	Flexion
(-) Negative	-	Extension
ASTM	-	<b>American Society for Testing and Materials</b>
BMT	-	<b>Bending Moment Transmitter</b>
°C	-	Degrees <b>Celsius</b>
DCP	-	<b>Dynamic Compression Plate</b>
Ens. Avg.	-	Ensemble Average
EZ	-	<b>Elastic Zone</b>
EZS	-	<b>Elastic Zone Stiffness</b>
F	-	<b>Female</b>
FBC-SLS	-	<b>Free Bending Canine Spinal Loading Simulator</b>
FBCS	-	<b>Free Bending Canine Spine</b>
GSD	-	<b>German Shepard Dog</b>
kN	-	<b>Kilo Newton</b>
L5	-	5 <sup>th</sup> lumbar vertebra
L6	-	6 <sup>th</sup> lumbar vertebra
L7	-	7 <sup>th</sup> lumbar vertebra
LC-DCP	-	<b>Limited Contact Dynamic Compression Plate</b>





M	-	<b>Male</b>
N.m	-	<b>Newton meter</b>
NZ	-	<b>Neutral Zone</b>
NZS	-	<b>Neutral Zone Stiffness</b>
ORIF	-	<b>Open Reduction with Internal Fixation</b>
PMMA	-	Polymethylmethacrylate bone cement
Potting	-	Fixation of the specimen in prefabricated cups (using epoxy) in preparation for mounting in Free Bending Canine Spinal Loading Simulator
RoM	-	<b>Range of Motion</b>
S1	-	1 <sup>st</sup> sacral vertebra
S2	-	2 <sup>nd</sup> sacral vertebra
SD	-	<b>Standard Deviation</b>
SLS	-	<b>Spinal Loading Simulator</b>
SOP	-	<b>String of Pearls</b> Interlocking Plate
SPCA	-	<b>Society for the Prevention of Cruelty to Animals</b>

# CHAPTER 1: INTRODUCTION

## 1.1 Background

Spinal injuries seen in small animal practice are usually associated with a severe traumatic event wherein the stabilising functions of normal structures are overwhelmed by excessive external flexional forces, most commonly as a result of motor vehicle trauma.

Vertebral fracture-luxation is a painful condition which is usually associated with varying degrees of neurological deficits due to neural compression or direct mechanical injury.

In some canine patients displaying mild neurological symptoms, conservative therapy consisting of strict cage confinement and pain control may be all that is required for treatment. A study by Selcer *et al.*, (1991) on the management of vertebral column fractures in dogs and cats ( $n=211$ ) showed very little difference in the final outcome of surgically treated patients versus medically treated patients. However, medically treated patients took up to eight months longer to reach optimal neurological outcome when compared to surgically treated patients.

The primary objectives of surgical treatment of spinal fracture-luxations include preventing any further damage to the spinal cord and nerve roots, relieving any compression present, and applying rigid stabilisation. Perren (1979) determined that less than 2% elongation should be present over the callus at the fractured bone ends in order for primary bone healing to occur<sup>31</sup>. The ideal fixation method must therefore be able to withstand all the disruptive forces generated by the muscles of the patient as well as the intrinsic and external biomechanical forces acting on the fracture site during movement. Connective tissue in the external or periosteal callus can withstand 80% elongation and fibrocartilage tissue in the callus can withstand up to 17% elongation before failure<sup>31</sup>. Any excessive instability over a

fracture line, especially during the early stages of bone healing, will therefore result in delayed healing and excessive callus formation or even non-union.

In most dogs, the spinal cord terminates at the level of lumbar vertebra six (L6). Caudal to that area, the vertebral canal in the lumbosacral region contains the cauda equina rather than the spinal cord, with a spinal canal relatively spacious for the nerve roots that it encloses. Injuries to the lumbosacral region generally cause less severe neurological deficits (hind limb pain, reluctance to stand and walk, caudal lumbar pain, decreased anal tone, urinary retention, and decreased tail sensation) and usually have a better prognosis than injuries in other regions of the spinal column <sup>6; 19; 46; 51</sup>.

The surgical objective for treatment of a L7 fracture-luxation includes decompressing the cauda equina – which can usually be accomplished through fracture reduction – as well as adequately stabilising the spinal column.

In human patients, surgical intervention is not always used for the treatment of spinal fractures. However, studies have shown that rigid surgical stabilisation results in marked pain relief, improved mobility, and an enhanced healing process <sup>15</sup>.

Several methods for stabilisation of vertebrae have been described. Techniques range from the use of one or more transilial pins; screw fixation of the articular facets; vertebral body plating using conventional DCP plates, or locking plates of different configurations; pedicle screw-rod fixation; dorsal spinous process vertebral plating; vertebral stapling; vertebral spinous process pinning with wiring; pin or screws bonded together with polymethylmethacrylate bone cement (PMMA); as well as the use of external fixation and/or combinations of different techniques <sup>1, 4, 6-8, 10,13</sup>.

Two specific internal fixation methods are considered in this study: dorsal stabilisation using pins and polymethylmethacrylate (PMMA) <sup>7</sup>, referred to hereafter as pin-PMMA, and a

locking plate system called the String of Pearls Interlocking Plate system (Orthomed Ltd, Halifax, UK), referred to hereafter as SOP.

Dorsal stabilisation of spinal fractures using screws or pins and PMMA is a well-described technique, especially in the thoracic and lumbar spinal regions. SOP is a novel locking plate system that has been designed for use in veterinary and human orthopaedic surgery. Both techniques have several advantages and disadvantages which need to be taken into consideration when selecting the optimal fixation technique for the patient.

## 1.2 Problem statement

One of the biggest challenges in veterinary orthopaedic surgery, in particular large and giant breed dogs is restricting the patient's long-term post-operative activity. Therefore, knowledge of the biomechanical characteristics of the fixation method to be utilised for fracture stabilisation is of critical importance in order to make an informed decision regarding the use of a specific implant, as the strengths and weaknesses of the chosen method need to be taken into consideration. Patients also need to be handled and turned regularly to prevent cutaneous pressure sores, as they are unable to turn during the immediate postoperative period. The challenge of handling and treating large breed dogs is amplified by their increased body weight.

Walter *et al.*, (1986) proved that the centre of gravity of the caudal half of the dog (caudal to T13) is at the L7-S1 level<sup>44</sup>. Because of this, rigid internal fixation techniques are particularly important in this area, and careful handling is needed when lifting the paralysed animal by its chest in the clinical scenario.

Due to the increased risk of damage to the lumbosacral plexus, conventional dorso-lateral plating for fixation of the L7-S1 fracture-luxations is often avoided by surgeons. One advantage of the SOP locking plate system is the fact that the plate does not have to be lagged onto the underlying vertebral body's bone. Even though care must still be taken

during screw insertion to avoid trauma to the nerve roots, the likelihood of damaging the spinal nerves exiting the intervertebral foramina in this region is reduced, because the plate “stands off” from the vertebrae, rather than compressing against the bone<sup>12, 21</sup>.

The use of dorso-lateral plating for fixation of the L7-S1 fracture-luxation is technically more demanding and emphasises the importance of testing this new fixation method and comparing it with the more conventional methods which are currently used in practice.

Knowledge of the biomechanical characteristics, along with the surgical considerations of the stabilisation method, is of critical importance to make an informed decision regarding the selection of a specific method<sup>16</sup>.

### **1.3 Aim**

The aim of this study is to quantify and compare the stability of two internal spinal fixation techniques in flexion and extension when applied to a surgically simulated, complete spinal injury at the level of L7-S1.

The two fixation techniques of interest are:

- 1) Pins and polymethylmethacrylate (pin-PMMA) and
- 2) String of Pearls (SOP) locking plates

### **1.4 Hypothesis**

It is hypothesised that in the treatment of lumbosacral fracture–luxations of large breed dogs, instability fixated with SOP consisting of two bilateral SOP plates, anchored in the L6, L7, S1 and S2 vertebral bodies, will be equally as stable as the conventional pin-PMMA method using four positive profile end-threaded pins anchored in the vertebral bodies of L7 and S1 and bonded dorsally with PMMA.

## 1.5 Objective and value of this study

The primary objectives of this study are therefore to:

- Determine and compare the stability of the two fixation techniques at the level of the injured L7-S1 joint
- Determine and compare the static strength of the two fixation techniques

A secondary objective is to:

- Compare the behaviour of the spinal joints adjacent to the injured lumbosacral joint between the two fixation techniques

New implants and surgical approaches are continuously being developed, which can be overwhelming for a surgeon having to select the best implant for a specific surgical case. The type of implant used is often determined by the preference and experience of the surgeon, personal experience with the method, location of the fracture, and the cost of the implant – and may not always represent the best method for the situation.

Using standardised biomechanical laboratory tests to compare the different characteristics of an implant to more commonly used techniques will enable surgeons to make more informed decisions and to avoid potential implant-related complications<sup>26; 40; 50</sup>.

One of the aspects that the surgeon must consider is the stability of the fixation technique. The ideal fixation technique for vertebral fractures should be rigid enough to encourage normal bone healing and be able to withstand all the biomechanical forces on the spinal column (internal and external) during the rehabilitation process, while the animal is in normal ambulation and motion<sup>19</sup>. The strength of the implant or fixation technique is also of utmost

importance because the implant should be able to provide stability without failing under *in vivo* physiological loads.

No study was found in the literature which compares the stability during flexion and extension of SOP interlocking plates with more commonly used and proven methods of stabilisation, such as pin-PMMA – specifically when used for the stabilisation of lumbosacral fracture-luxations.

## **1.6 Conflict of interest**

None of the authors of this study have a financial or personal relationship with other people or organisations that could inappropriately influence or bias the findings of this study.

The Department Companion Animal Clinical Studies, Faculty of Veterinary Science, University of Pretoria, South Africa, supported part of the funding for this study.

The funders had no role in study design, data collection and analysis, decision to publish, or preparation of the manuscript.

## CHAPTER 2: LITERATURE REVIEW

### 2.1 Anatomy and Biomechanical considerations

The lumbar segment of the canine vertebral column consists of seven vertebrae, each one gradually increasing in length caudally up to the sixth lumbar vertebrae (L6). The length of the seventh lumbar vertebra (L7) is approximately the same as the first lumbar vertebra (L1)<sup>45</sup>. The articulation facets are orientated in the sagittal plane; allowing flexion and extension of the lumbar spinal column, but restricting lateral bending to a large degree.

A dog's lumbar vertebrae have large articulation facets and accessory processes, with several surrounding ligaments creating a delicate balance between stability and mobility.

The sacrum of the dog is formed by the bony fusion of three sacral vertebrae with the cranial base and the caudal apex; highly mobile articulations are formed with L7 and the first caudal vertebrae respectively.

In most dogs, the spinal cord decreases rapidly in size – caudal to lumbar vertebra four (L4) in non-chondrodystrophic breeds, and caudal to lumbar vertebra five (L5) in chondrodystrophic breeds – and terminates at the level of lumbar vertebra six (L6).

The vertebral canal in the lumbosacral region contains the cauda equina rather than the spinal cord, and the spinal canal in this region is relatively spacious for the nerve roots that it encloses.

The dorsal and ventral branches of the first and second sacral nerves exit the sacrum through dorsal and ventral foramina. The last sacral spinal nerves exit between the sacrum and the first caudal vertebra.

Intervertebral discs between individual vertebrae provide flexibility to the vertebral column, function as shock absorbers for the spine, and are the single most important stabilising



factor against rotational forces acting on the spine. The lumbosacral disc is the largest intervertebral disc in the canine spinal column<sup>24</sup>.

The canine lumbar spine allows movement in three dimensions: flexion and extension with slight lateral bending, and axial rotation. Anatomically, the lumbar spine is best suited to withstand forces in flexion and extension, both of which are important components of normal canine locomotion<sup>8</sup>.

The very strong epaxial and hypaxial muscles associated with the lumbar area restrict movement largely to the sagittal plane, with high resistance to torsional forces. A study by Benninger *et al.*, (2004) showed that when flexion and extension are the main motion, the effects of lateral bending and rotation are actually minimal<sup>7</sup>.

The forces generated during propulsion from the pelvic limbs are transmitted via the caudal lumbar vertebrae and cause traction and compression of the vertebrae – which explains the typical fracture-luxations seen in this area (see Figure 2.1).

However, most veterinary biomechanical evaluations of spinal fixation methods have focused on the model's stability in flexion and extension with the ultimate strength tested in flexion<sup>10,37</sup>. Therefore, in this research study, flexion was evaluated under both physiological loads and eventually to failure.

## 2.2 Spinal fractures

Lumbosacral fracture–luxation most often results from failure of the L7 caudal-ventral vertebral buttress, articulation facets, and the pedicles; while most of the L7 vertebral body remains intact. Lumbosacral fracture–luxations have a very characteristic radiographic appearance – a small oblique or wedge-shaped fracture – and extend from the lumbosacral intervertebral foramen through the caudal body of L7 vertebral body<sup>10</sup>. (Figure 2.1)

During lumbosacral fracture-luxation, the muscular forces acting on the sacrum and pelvis, along with the weight of the dog's pelvic mass, typically cause cranio-ventral displacement of the caudal vertebral body segment.

A diagnosis of vertebral fracture-luxation can generally be made using radiographs. However, some cases may show typical clinical signs, but the fracture luxation cannot be demonstrated with radiographs. These isolated cases may require computed tomography (CT) or magnetic resonance imaging (MRI) <sup>45:49</sup>.

Seim *et al.*, (2002) suggest that the lumbosacral joint is prone to injury due to the static-kinetic relationship between the mobile caudal lumbar vertebral bodies and the relatively fixed sacrum. The high level of mobility in this area, where all the forces from the hind limbs are transmitted to the caudal lumbar vertebrae via the pelvis and sacro-iliac joint, makes the caudal lumbar area more prone to disc degeneration and trauma <sup>35</sup>.



**Figure 2.1:** Lateral radiograph of the caudal lumbar spine, illustrating a typical lumbosacral fracture–luxation. A small oblique fracture of the caudal body of L7, with typical cranio-ventral displacement of the caudal fragment and sacrum, is clearly visible.

Nerve roots can tolerate more deformation and recover better from concussive damage than the spinal cord neuro-parenchyma, and they usually carry a better prognosis compared to injuries in other regions of the spinal column<sup>6; 10; 35; 41; 46; 48</sup>.

The prognosis is therefore based on the neurological symptoms, not on the radiological findings, because even substantial displacement of the fracture or luxation may still leave the patient neurologically intact<sup>41; 46</sup>. An important prognostic indicator is the ability to perceive deep nociception<sup>51</sup>.

Fracture–luxation of L7 can, however, traumatise the nerve roots of the cauda equina (L6, L7, and sacral nerve roots). Clinical signs frequently observed are hind limb pain, reluctance to stand and walk, pain and crepitation over the caudal lumbar area, decreased anal tone, urinary retention, and decreased tail sensation<sup>6; 19; 46; 51</sup>.

Decreased conscious proprioception of the hind limbs is usually present, but may be difficult to evaluate because the patient is reluctant to support any weight on the pelvic limbs<sup>41</sup>.

## 2.3 Treatment of Fractures

Surgical stabilisation of L7 fractures is challenging due to the potential for iatrogenic damage to the lumbosacral nerve plexus during surgery<sup>6</sup>.

Because the fracture luxation of L7 is usually associated with a severe traumatic incident, such as a motor vehicle accident, moderate to severe concomitant injuries are frequently present. These will require intensive treatment and stabilisation before any surgical correction is attempted.

In human medicine, the timing of decompressive spinal surgery is controversial because spinal trauma patients often have concurrent multiple organ traumas which can make early intervention more risky.

The current tendency, however, is towards early intervention (within 24 hours) and internal stabilisation. This is associated with statistically better outcomes compared to both delayed decompression and conservative treatment <sup>15</sup>.

In some canine patients showing mild neurological symptoms, conservative therapy may be all that is required for treatment <sup>13</sup>. In these cases, the surrounding musculature and the remaining inherent stability of the vertebral column may prevent further movement and trauma to the nerve roots.

A major problem for the veterinarian is the challenge of immobilising and managing spinal fracture cases which are treated conservatively for prolonged periods of time. External bracing or casting does help to a certain degree, but is often not sufficient to provide adequate support and stability – especially in larger dog breeds <sup>8; 23</sup>. It is therefore easier to care for patients for a relatively shorter period after surgical fixation of a fracture luxation, than the extended period required for patients treated medically with cumbersome external coaptation devices <sup>19</sup>.

Additionally, rapid improvement of neurological function is usually seen after rigid, internal surgical fracture stabilisation <sup>41</sup>.

As stated before, the objectives of surgical treatment of spinal fracture-luxations are to prevent further damage to the nerve roots and to relieve any compression present. Decompression is usually adequately achieved by reducing the fracture luxation. The spinal cord and the cauda equine in particular, can tolerate some compression; provided that the compression is static. Near-perfect reduction with good stability may be of greater importance than perfect alignment, which is potentially associated with additional iatrogenic trauma <sup>19</sup>.

Until bone healing has occurred, the fixation technique used must be able to withstand all the disruptive physiological forces generated by the muscles of the patient itself during

movement, as well as all the other forces placed over the fracture line during the handling of the animal <sup>23</sup>.

The strain theory of bone addresses the mechanism by which a fracture will heal – primary, secondary, or non-union <sup>31</sup>. Bone healing strain is expressed as a relationship between the amounts of displacement of the bone fragments at the fracture site divided by the fracture length. In order for primary bone healing to occur, less than 2% elongation must be present over the callus at the fracture bone ends. Where elongation between 2-10% is present, secondary bone healing with callus formation takes place. In cases where elongation exceeds 10%, bone resorption and even non-union may occur <sup>18; 27; 40</sup>. Therefore, instability across the fracture line will result in delayed bone healing with excessive callus formation or even non-union. The ideal fixation methods should be able to withstand minimal angular deformation.

The vertebral bodies consist mainly of well-vascularised cancellous bone and, once reduced and rigidly stabilised, usually heal rapidly with minimal callus formation <sup>19</sup>.

Another important goal for fracture fixation is early ambulation after surgery – something which is rarely achieved with the use of conservative methods of treatment. Even though fracture healing using open reduction with internal fixation (ORIF) methods can take, on average, up to five times longer to heal, the implant protects the fracture and therefore allows physiological loading with early mobilisation <sup>27; 32</sup>.

## **2.4 Methods for stabilising vertebral fracture-luxations**

In veterinary science, the two key methods used for stabilising vertebral fracture-luxations in dogs are pins bonded with PMMA, or bone plates – either conventional or locking plates with screws. Both of these methods have advantages and disadvantages <sup>24</sup>.

## 2.4.1 Pin and PMMA

Dorsal stabilisation of spinal fractures using PMMA and smooth or threaded pins is a well-described technique that is commonly used in the treatment of thoracic and lumbar spinal fracture stabilisation. Due to the lower cost of pins and PMMA and the minimal instrumentation needed, this technique is popular in veterinary surgery <sup>6; 8; 16; 43; 44; 46; 51</sup>.

Replacing the pins with bone screws – bonded dorsally with PMMA – has also been used for lumbosacral fracture fixation with fairly good results <sup>6; 38</sup>.

The principle of using pin-PMMA for fracture-luxation stabilisation is to insert the pins – preferably positive profile – into the vertebral bodies, cranial and caudal to the luxated or fractured vertebra. The pins are then notched, bent, and bonded dorsally with PMMA to create a rigid internal fixation system <sup>46</sup>. The strength of the pin-PMMA construct is strongly influenced by the diameter of the pins used <sup>6; 16; 46</sup>.

The formula for the resistance of a pin to bending is calculated as follows:

$$\sigma = \frac{\pi r^4}{4}$$

where  $\sigma$  is the bending resistance and  $r = 0.5$  times the core diameter of the pin.

From this equation, it is clear that a quartic relationship between the resistance to pin bending and the pin's radius exists, and a small increase in pin diameter will thus greatly increase its resistance to bending <sup>6</sup>. Therefore, the largest diameter pin as possible should be used as implant, based on the size of the patient <sup>46</sup>.

Blass *et al.*, (1986) suggested that the volume of PMMA used should be 20 grams in dogs weighing less than 15 kg and 40-60 grams in larger dogs <sup>8</sup>. However, Beaver *et al.*, (1996) concluded that the diameter of the PMMA cylinder is more critical than the quantity used <sup>6</sup>.

It is important to embed the dorsal aspects of the pins in the PMMA to reduce the change of pin migration <sup>8</sup>.

The pin-PMMA method of stabilisation is flexible in terms of the number and position of the pins used, can be applied at most levels of the vertebral column, and it is resistant to rotational forces <sup>24</sup>. Another advantage of using pin-PMMA is the fact that it accommodates patients of a large variety of body weights, as the pin size and the volume of PMMA needed to bond the pins can be adjusted to suit each patient <sup>51</sup>. This method also avoids the possibility of compressing the spinal nerve roots <sup>24</sup>. Additionally, less soft tissue dissection is normally required for placement compared to most other methods, and a shorter segment of vertebral column needs to be immobilised, allowing for more normal spinal function <sup>6</sup>.

An external skeletal fixator construct, using a 19 mm tubular column of PMMA, compares favourably with constructs using 4.8 mm stainless steel bars <sup>4</sup>.

The main disadvantages of using the pin-PMMA method are thermal injury due to the exothermic reaction during the setting of the PMMA, pin migration, and also increased risk of seroma formation and infection in the area <sup>23; 24</sup>. The bulk of the implant also makes closure of the wound more difficult <sup>49</sup>.

The problem of pin migration or pin pull-out can be avoided or minimised with the use of partially threaded – preferably positive-profile pins, instead of smooth pins <sup>6; 51</sup>. Positive-profile threaded pins are partially threaded pins with a thread diameter bigger than the core diameter of the pin. This reduces the stress at the bone-pin interface, reducing the risk of pin breakage and, to a large degree, the possibility of pin migration <sup>51</sup>.

In the clinical situation, antibiotics are incorporated into the polymer powder used in an attempt to counter latent infections associated with PMMA <sup>6</sup>.

Four positive-threaded Kirschner pins are inserted in the adjacent luxated vertebrae, or cranial and caudal to the fracture line of a fractured vertebra. To optimise pin-bone contact,

the pin is driven at an angle, optimally penetrating the far cortex (ventrally) a few millimetres. In order to avoid damage to critical structures such as the aorta, care is taken not to drive the pins too far beyond the trans-cortex.

The pins are notched and bent at the level of the dorsal spinous processes, and bonded together with PMMA to create a rigid internal fixation system<sup>43</sup>. The heat created by the polymerisation of the PMMA is dissipated by flushing the area with cool lactated Ringers or sterile saline (0.9% NaCl) solution.

Published biomechanical studies and clinical reports describe the use of up to eight or ten screws or pins in the lumbar vertebrae, immobilising two or even three intervertebral disc spaces. The iliac wing can also be incorporated into the fixation and bonded with PMMA<sup>6; 16</sup>. These methods of fixation, however, require extensive surgical exposure when compared to the conventional four pin fixation strategies in other areas of the vertebral column. Walker *et al.*, (2002) showed in a biomechanical study that four pin-PMMA constructs used in a canine lumbar spinal model were just as stiff as intact spines in flexion and in extension<sup>43</sup>. In the clinical situation, vertebral body fixation techniques incorporating as few vertebrae as possible, seem to be superior to dorsal fixation techniques spanning several vertebrae. The fixation of only one vertebra cranial and caudal to the fracture-luxation reduces the forces that are transmitted through the fixation device during the flexion and extension of the spine<sup>51</sup>.

The pedicle of L7 is larger compared to the other lumbar vertebrae, allowing screw and/or pin placement into the pedicle as well as into the vertebral body in the sagittal plane – while avoiding interference with the iliac wing during implant placement<sup>45</sup>.

Méheust *et al.*, (2000), using pedicle screw fixation to stabilise the lumbosacral area, determined that the ideal point of entry into the L7 vertebral body is located at the small bony crest just caudal to the base of the cranial articulation facet and parallel to the sagittal plane (see Figure 2.6)<sup>25</sup>. For S1, the ideal point of entry is at the sloped plane just caudo-lateral to



the cranial articulation facet, with the screw placed parallel to the vertebral end plate of the sacrum (see Figure 2.7) <sup>25</sup>.

Weh *et al.*, (2007) used the same entry point in L7 caudal to the base of the cranial articulation facet, and directed the pins cranio-ventro-medially to exit the vertebral body of L7 cranio-ventrally. This allowed maximum purchase of the pins in its pedicle and vertebral body while avoiding the fracture site <sup>46</sup>.

Lumbosacral fracture-luxations mostly result from failure of the caudal-ventral vertebral buttress of L7 while the rest of the L7 vertebra still remains intact <sup>46</sup>.

Provided that the cranial fragment is large enough and the fracture is not highly comminuted, these fracture-luxations can be stabilised with cranial implants placed into the pedicles and vertebral body of L7, and the caudal implants into the sacrum. If a comminuted fracture of L7 is present, the cranial implants can be placed in L6, spanning L7 <sup>25; 46</sup>.

Up to now, biomechanical studies have not yet confirmed an ideal configuration for pin-PMMA constructs <sup>16; 43; 44</sup>.

## 2.4.2 Vertebral body plating

Conventional vertebral body plating shows good resistance to flexion-extension forces, but less resistance to rotational forces, requiring accurate plate contouring following the reduction of the vertebral body <sup>24</sup>.

The application of conventional plates can be technically demanding due to the relative size of the plate versus the vertebral body size. This often makes it quite difficult to place the correct amount of screws cranially and caudally to the fracture line, and the ilium wings may interfere with access to the vertebral bodies <sup>19</sup>.

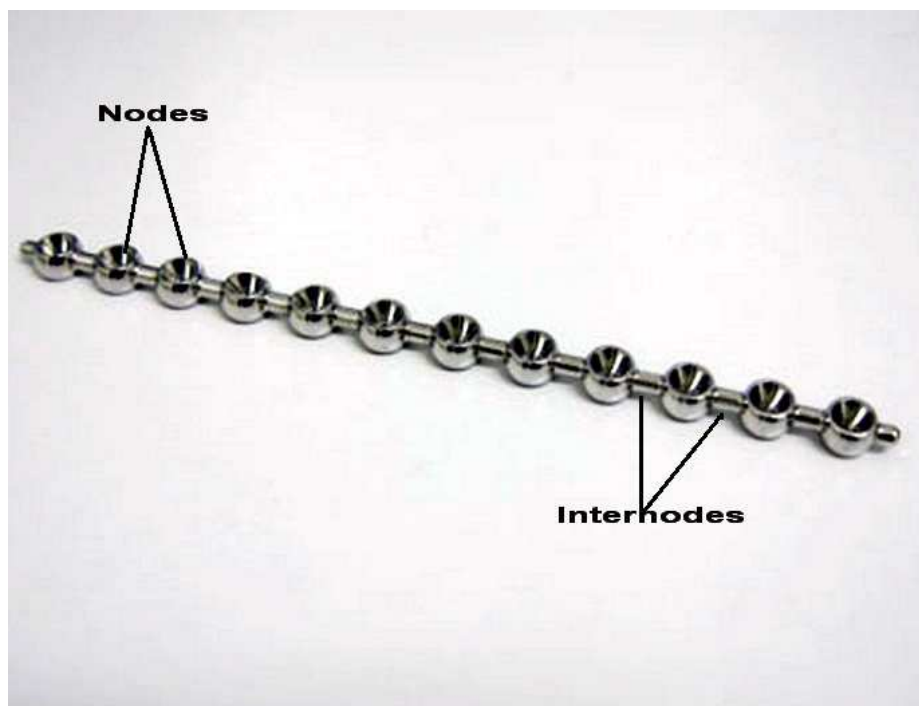
Screws in conventional bone plates press the plate onto the underlying bone when the screws are tightened, relying on friction generated between the plate and bone and between the screw head and the plate to ensure stability <sup>2; 9; 11; 21; 22; 27; 30; 40; 42</sup>.

Locking plates, however, have evolved over the last few years in an effort to overcome some of the limitations associated with using conventional bone plates, and are especially useful for fixation of bone that has poor quality – such as osteopaenic bone<sup>40</sup>.

The biomechanics of interlocking plate systems differ fundamentally from that of the conventional bone plates<sup>22</sup>.

### 2.4.3 SOP locking Plates

The String of Pearls (SOP) Interlocking Plate system (Orthomed Ltd, Halifax, UK) is a locking plate that has been used successfully to provide adequate stabilisation when treating spinal fractures in small animal patients<sup>1; 12; 21; 22; 24</sup>. The SOP system is machined from 316 LVM surgical stainless steel wrought bar stock, and is manufactured according to the American Society for Testing and Materials specifications (ASTM F138-03)<sup>30</sup>. The SOP system consists of a series of cylindrical sections, called internodes, and spherical components, called pearls or nodes (see Figure 2.2).



**Figure 2.2:** Close-up view of a String of Pearls (SOP) Interlocking Plate system, without screws inserted, consisting of a series of internodes and spherical components (nodes).

The 3.5 mm SOP plate's internodes have a diameter of 5.00 mm and the nodes are spheres with an 8.00 mm diameter. The top of the node is wider, allowing a standard cortical screw to recede into the plate when tightened. The base of the node portion contains a threaded portion. As the screw head recedes into the node, it makes contact with a ridge inside the node. This causes a press fit of the screw into the node, which prevents the screw from loosening during cyclic loading of the implant<sup>22; 28</sup> (see Figure 2.3).



**Figure 2.3:** Close-up view of a SOP Interlocking Plate. The top of the node is wider, allowing a standard cortical screw to recede into the plate when tightened. The base of the node portion contains a threaded portion, causing a press fit of the screw in the node.

The node accepts a drill/tap guide; allowing drilling, measuring with a depth gauge, and tapping the screw hole using standard ORIF instrumentation.

The increased diameter of the nodes relative to the internodes gives the implant a relatively consistent stiffness profile, therefore the screw holes are not weak points as in a conventional bone plate<sup>22</sup>.

Mechanical testing using ASTM standards demonstrated that the 3.5 mm SOP plate is approximately 50% stiffer, and has a bending strength – the load at which the plate plastically deforms or bends – of 16% to 30% greater than the 3.5 mm Limited Contact Plate

(LCP), Dynamic Compression Plate (DCP), or Limited Contact – Dynamic Compression Plate (LC-DCP) <sup>21; 22; 28; 29</sup> .

Twisting the SOP plate does weaken it slightly, but mechanical testing using ASTM standards has proven that, although bending and twisting the SOP plate may reduce its stiffness and strength by about 33%, a 3.5 mm SOP plate being bent through 40° remains 96% as stiff as a new, untouched 3.5 mm DCP plate <sup>29</sup> .

However, severe contouring can cause deformation of the screw holes, which may prevent the screws from locking securely into the nodes. Plate deformation can be prevented by inserting small metallic inserts, called bending tees, into each hole of the SOP plate prior to contouring with the SOP bending tools (Orthomed UK Ltd, Halifax, UK) (see Figure 2.4) <sup>22; 29</sup> .



**Figure 2.4:** The use of bending tees (see above), inserted into each hole of the SOP plate prior to contouring the plate, will prevent deformation and retain the plate's locking function. The photo shows four tees inserted into the SOP plate and three separate tees.

Advantages of the SOP system include the use of standard cortical bone screws as locking screws; the ability to contour the plate into six degrees of freedom (two plane bending and

torsion); high bending strength; and very competitive pricing, as costing is currently similar or less than that of conventional orthopaedic plating systems <sup>21; 22; 29</sup> .

Due to their ability to contour the plate into six degrees of freedom, these SOP plates can be accurately contoured when applied to the dorso-lateral aspect of the thoracic and lumbar spinal vertebrae <sup>22; 24</sup> .

Like all locking plate systems, the SOP also functions as a small internal skeletal fixator <sup>2; 40</sup> . Due to the relatively close proximity of the plate to the bone, the locked screw lengths are significantly shorter than conventional external fixator pins, contributing to the mechanical rigidity of the locking plate system <sup>40</sup> . External fixation is used for surgical treatment of lumbosacral instability as seen secondary to discospondylitis, but the post-operative management usually proves a challenge for the client and the patient <sup>2; 5</sup> . Locking plates, such as the SOP system, therefore combine the advantages of external skeletal fixation with the relative ease of management of an internal fixation system <sup>2; 11</sup> .

During normal weight bearing, the primary loads on the underlying bone are axial, running along the long axis of the bone <sup>22</sup> . After fracture fixation with a conventional plate and screws, the axial loads on the bone encounter a screw and the load is transferred at the bone/screw interface: first to the screw, then to the plate, then to the screw on the other side of the fracture, and finally back to the bone again <sup>21; 22</sup> .

Screws in conventional bone plates press the plate onto the underlying bone at a force of about 2-3 kilo-Newton (kN) as the screws are tightened <sup>33</sup> . Conventional plating systems rely on the friction that is generated between the plate and the bone, and between the screw head and the plate to ensure stability <sup>9; 11; 27; 28; 40</sup> . This friction is generated by the screws pressing the plate onto the bone's surface. Therefore, the ability of a screw to resist pull-out is crucial to the functioning of conventional plating systems <sup>20</sup> .

In contrast, the screws of interlocking plates are firmly fixed to the plate; eliminating screw toggle and turning the plate/screw combination into a fixed-angle, single-beam construct <sup>40</sup> .

The plate is not being pressed down onto the bone, so the screw's resistance to pull-out is less relevant<sup>22; 42</sup>. The screws act as transverse supporting structures which are subjected to cantilever bending, so factors other than the interdigitation between the screw threads and the bone contribute to security and stability. The screw is always an integral part of the transmission of forces across the specific areas of the bone fracture<sup>24; 40</sup>.

The stability of the locking plate across the fracture site depends on the load applied as well as the mechanical properties of the plate and screws. Mechanical properties which are important are the length and cross section of the plate, material properties of the plate, use of monocortical versus bi-cortical screws, and the diameter of the screws<sup>40</sup>.

The fatigue life of the screw/plate interface increases dramatically in importance<sup>22</sup>. A very significant mechanical feature of all interlocking plate systems, including the SOP, is the fact that there is a distinct stress riser at the screw/plate interface, where the forces are transferred from a less stiff element, the screw, to a much stiffer element, the plate. Excessive forces that are applied cyclically across the fracture line will cold work the shaft of the screw and cause the screw to become more brittle, cracking and eventually breaking over time<sup>22</sup>.

Therefore, when using the SOP system it is of relatively less importance to engage two cortices with each bone screw, and of much greater importance to rather increase the number of bone screws, uni-cortical or bi-cortical, to enhance the screw's fatigue life. By adding another uni-cortical screw may have a limited benefit with conventional bone plates, but uni-cortical screws within a locking plate system will enhance the fatigue life of the locking construct<sup>21; 22; 42</sup>.

Based on theoretical considerations, it is suggested that at least three– and preferably four screws should be used for each major bone fragment in order to protect the screws against early metal fatigue and premature failure<sup>22</sup>.

The cross-sectional area ( $A$ ) of the 3.5 mm SOP is  $A = \pi r^2$  or 20 mm<sup>2</sup>, and the cross-sectional area of the screw is about 5 mm<sup>2</sup>. Therefore, by installing four screws on either side of the fracture, the shear area of the four screws will be approximately equal to that of the SOP plate, subsequently protecting the screws against premature fatigue failure<sup>22</sup>.

A second SOP plate can be used on the contralateral or orthogonal side of the spinal column, or two SOP's can be nestled side by side. The use of SOP plates in pairs in the fixation of spinal column fractures should be considered the norm<sup>22</sup>.

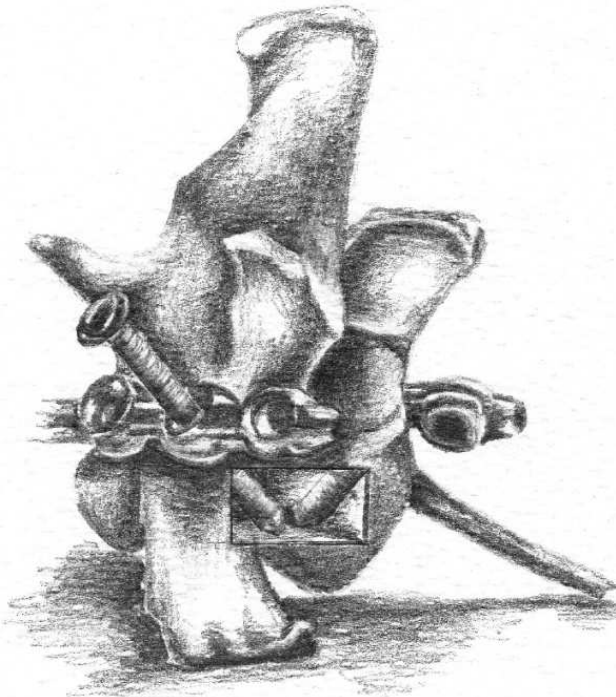
Conventional plating systems that are applied to weak or osteopaenic bones mostly fail due to the screws pulling out of the bone prematurely. In locking plate systems, the fact that the screw is locked in the plate prevents the screws from pulling out so easily. Instead, failure usually occurs through the slow creep of the screw through the weak bone – also known as “bone slicing”<sup>22; 28</sup>.

An advantage of the SOP locking plate system, when used for fixation of spinal fractures, is that the plate does not have to be lagged directly onto the underlying vertebral body bone, thereby accommodating irregularities of the vertebral column. The likelihood of damaging spinal nerves exiting the intervertebral foramina is now also reduced, because the plate “stands off” from the vertebrae, rather than compressing against it<sup>12; 21</sup>.

Ahmad *et al.*, (2007) looked at the issue of clearance of the LCP and used the DCP as a comparison<sup>3</sup>. LCP plates were either placed flush against the bone, similar to a DCP plate, or placed 2 mm to 5 mm away from the bone by using three locking screws into each bone fragment. Results showed that the LCP that was placed flush against the bone and the LCP that was placed 2 mm away from of the bone responded similarly to a DCP plate and failed at a high load. However, the LCP placed 5 mm from the bone failed at a significantly lower average static loading. It was concluded that even though it is crucial to preserve the periosteal blood supply, it is advised that the distance between the bone and plate should preferably not exceed 2 mm.

The fact that torsion can be applied to the SOP plate makes some variation in screw angle possible and indicates that the plate is very versatile for use in many orthopaedic procedures, such as acetabular fractures. However, accurate pre-contouring is essential, especially when used on the spine, because, unlike DCP plates where the screw angle can be varied to some degree, the locking plate screw design determines that the screw can only be placed perpendicular to the spherical node of the SOP plate <sup>12; 27</sup>.

Bilateral SOP plates are applied on the dorso-lateral aspect of the spinal column, with the screws directed at an angle of 60° (range 55°–65°) from the mid-sagittal plane into the lumbar vertebral bodies L1 to L6 (see Figure 2.5) <sup>43; 45</sup>.

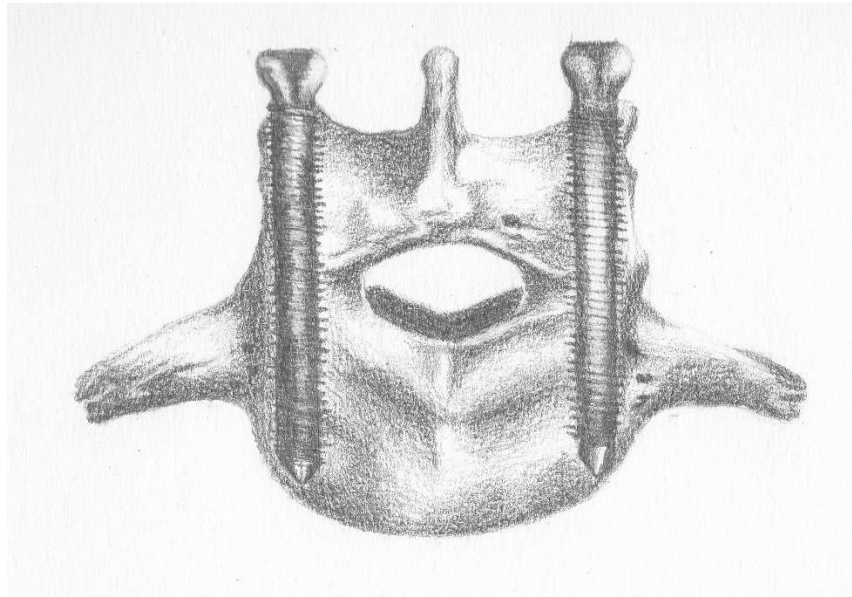


**Figure 2.5:** Left cranio-lateral view of the L6 vertebral body showing dorso-lateral applied bilateral SOP plates on the lateral aspect of the spine. The screws are directed at a prescribed angle of 60° from the mid-sagittal plane into the lumbar vertebral body. The cranial view has been “cut away” to show the correct screw placement. (Source: L. Liebenberg; Fourways Veterinary Hospital, Bryanston, South Africa)

The close proximity of the iliac wings makes implant placement in the vertebral body of L7 more difficult <sup>47</sup>. However, the pedicle of the L7 vertebra is wide enough to allow vertically directed implantation of pins or screws <sup>25; 45</sup>. The entry point of the implantation corridor is at

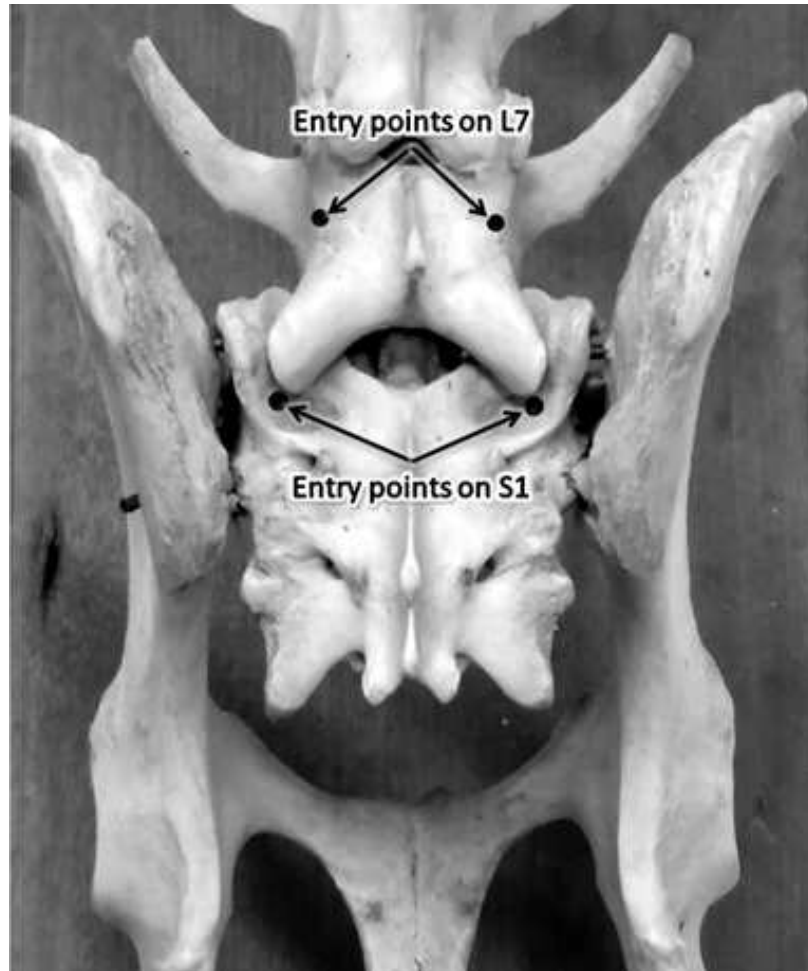


the base of the L7 cranial articulation process, with the screw orientated in the sagittal plane<sup>45</sup>. (Figure 2.6)



**Figure 2.6:** Cranial view of L7, showing recommended screw placement angle – the cranial end of the vertebra is cut away to show the screws. The entry point of the implantation corridor is at the base of the L7 cranial articulation process with the screw orientated in the sagittal plane.  
(Source: L. Liebenberg; Fourways Veterinary Hospital, Bryanston, South Africa)

As for L7, pedicle implantation is possible for S1. The entry point for S1 is located just caudal to the cranial articulation surface; with a mean laterally directed angle of 5° relative to the sagittal plane<sup>45</sup> (see Figure 2.7). In general, screw placement into S2 is not commonly used in the literature due to possible damage of the nerve roots. However, Coetzee (G.L. Coetzee, University of Pretoria, South Africa, personal communication, 2014) proposes – and regularly uses – placement of bilateral screws into S2. The SOP screw hole into S2 is drilled by using a smooth 2.5 mm Kirschner pin, instead of a 2.5 mm drill bit. No neurological deficits were noted in several cases done using this method.



**Figure 2.7:** Dorsal view of L6, L7, and sacrum depicting entry points for screw or pin placement. Méheust *et al.*, (2000) determined the ideal point of entry in L7 is located at the small bony crest at the base of the cranial articular facet, placed parallel to the sagittal plane. For S1, the ideal point of entry is at the sloped plane just caudal to the cranial articular facet, with the pin placed parallel to the vertebral end plate of the sacrum

25

## CHAPTER 3: MATERIALS AND METHODS

### 3.1 Introduction

Lumbosacral fracture-luxations are most commonly seen in younger dogs as a result of motor vehicle trauma<sup>35</sup>. Since any instability over a fracture line will result in delayed healing and excessive callus formation, the ideal fixation methods should be able to withstand even minimal angular deformation. Less than 2% elongation must be present at the callus and the fractured bone ends for optimal bone healing to take place<sup>31</sup>.

SOP plates are being used in practice for stabilisation of lumbosacral fracture-luxations, but no study was found in the literature that compares the stability of SOP plates during flexion and extension to more commonly used and proven methods such as the pins-PMMA for stabilisation of lumbosacral fracture-luxations.

The hypothesis is that, in the treatment of lumbosacral fracture–luxations of large breed dogs, lumbosacral instability fixated with SOP consisting of two bilateral SOP plates anchored in the L6, L7, S1, and S2 vertebral bodies, will be as stable as the conventional pin-PMMA method using four positive profile end-threaded pins anchored in the vertebral bodies of L7 and S1 and bonded dorsally with PMMA.

Although the study of flexion-extension load testing is informative, it cannot be regarded as complete. The canine spine is anatomically best suited to withstand forces in flexion and in extension<sup>37</sup>. A study by Benninger *et al.*, (2004) showed that when flexion and extension were the main motions, the effects of lateral bending and rotation were actually minimal<sup>7</sup>.

Most veterinary biomechanical evaluations of spinal fixation methods have focused on the model's stability in flexion as well as in extension, with the ultimate strength tested in flexion<sup>16; 36</sup>. Therefore, in this research project, flexion and extension will be evaluated under physiological loads as well as under increased loads up to eventual failure in flexion.

The terminology used in veterinary science of flexion and extension, can be confusing for people in the engineering field and was noticed during discussions. It happens frequently that different disciplines use different terminologies to describe the same concept.

The use of the word “flexion” to imply negative bending of the spine and the word “extension” to imply positive bending of the spine is strictly speaking incorrect from an engineering point of view. In mechanical engineering texts, “flexion” means “bending” and the chosen sign convention distinguishes between positive and negative. “Extension” in mechanical engineering texts means a purely axial loading that lengthens the sample under consideration. This can potentially be confusing when the two disciplines of engineering and veterinary science are working together as was noted during this research project.

For the purpose of this research the terminology as used commonly in veterinary science will therefore be used, where flexion means an upwards curvature of the spinal column and extension means a downwards curvature of the same spinal column.

The protocol was approved by the University of Pretoria Animal Ethics Committee (see appendix 2). All biological material was cremated after completion of the research project (*Envirocin Pet Cremation*; Kya Sands; Randburg; South Africa) as stipulated in the letter of approval from the National Society for the Prevention of Cruelty to Animals (NSPCA) (see Appendix 3).

### **3.2 Test Objective**

The objectives of the tests are to quantify and compare the strength and stability of two internal spinal fixation techniques in flexion and extension when applied to a surgically simulated complete spinal injury at the level of L7-S1.

The two fixation techniques are:

- 1) Pins and polymethylmethacrylate (PMMA) and,
- 2) String of Pearls (SOP) locking plates



### 3.3 Test methodology

The stability of the two internal spinal fixation techniques will be quantified by measuring the bending moment-angle characteristic of the joint with the surgically simulated complete spinal fracture-luxation – in this case, through the caudal aspect of L7 – in flexion and in extension within the elastic RoM of the spine segment. The spine segment will consist of the 5<sup>th</sup> lumbar vertebra, including the sacral vertebra (L5-S3). From the bending moment-angle characteristic, the biomechanical parameters will be extracted which quantify the stability of the spinal implant<sup>50</sup>. The bending moment-angle characteristic and the biomechanical parameters can then be used to compare the stability of the fixation techniques<sup>50</sup>.

The strength of the two internal spinal fixation techniques will be quantified by the load sustained at failure in flexion. Failure of the fixation technique will be defined as any catastrophic failure of the implant; that is, the abrupt loss of ability to support the load.

The load sustained at failure of the two fixation techniques will be used to compare the strength of the two techniques.

### 3.4 Specimens

#### 3.4.1 Inclusion criteria

The lumbar spine of 18 skeletally mature large breed dogs of both genders ( $29.84 \pm 2.49$  kg, mean  $\pm$  1SD) were used for this study. The dogs were euthanized as part of the SPCA program of humane euthanasia of unwanted animals over a two month period.

The animals were humanely killed by an intravenous overdose of sodium pentobarbitone (Eutha-naze<sup>®</sup>, Bayer (Pty) Ltd., Animal Health Division, Isando, South Africa). The dogs had no known history indicative of spinal cord or vertebral column disease or trauma, and radiographs were taken of the lumbosacral region to confirm the absence of any degenerative lumbosacral pathology or transitional vertebrae. CT was not performed in this

study. Uniformity in the cadaver material was kept as far as possible, but breed variation did occur due to availability of cadaver specimens at the local humane societies (see Table 3.1)

### 3.4.2 Initial specimen preparation

Within 2 hours after death, the lumbosacral spines were harvested. The segments that were harvested from the cadavers included the L3–L7 spinal bony segments, the sacrum with the pelvis attached, and the tail base with the first caudal (Cd) vertebra. These segments were cleared of most muscles; except for a thin layer covering the intervertebral articulations and dorsal interspinous ligament. All ligamentous tissues were left intact, but the pelvic canal tissue was removed. The pelvis containing the L3, L4, and the Cd vertebrae was removed, leaving the desired specimen, including only L5–S3. The spinous process and both transverse processes of L5 in each specimen were removed using a hacksaw to facilitate potting of the specimens in the epoxy (see Figure 3.1).



**Figure 3.1:** Dorsal view of typical test specimen (L5–S3), before potting cranial and caudal ends in epoxy. The spinous process and both transverse processes of L5 were removed to facilitate potting of the specimens in the epoxy. The cranial end (L5) is to the left in this image.

### 3.4.3 Identification of the specimens

Each specimen was individually and clearly marked with an identification number, breed, gender, and weight (see Table 3.1).

Identification	Breed	Gender	Weight [kg]
s1	GSD X	M	27.8
s2	GSD	M	27.4
s3	GSD	M	27.7
s4	GSD	F	26.9
s5	GSD	M	26.8
s6	Boerboel	M	30.1
s7	Boerboel	M	32.2
s8	Boerboel	M	34
s9	GSD	M	32.3
s10	Rottweiler	M	30.5
s11	Gt. Dane	M	30.6
s12	GSD	M	30.4
s13	Labrador	M	30.2
s14	Labrador	F	30.2
s15	Boerboel	M	35
s16	Boerboel	F	26
s17	Labrador	M	29
s18	Boerboel	M	30

**Table 3.1:** Identification of the specimens. Specimens were individually marked according to identification number, breed, gender and weight. (GSD – German Shepherd Dog; M – male; F – female)

### 3.4.4 Storage of specimens

Specimens were individually wrapped in cotton towels, soaked in lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg), and sealed

in marked plastic bags (Ziplock®, S.C. Johnson & Son). All specimens were stored in a domestic freezer (LG® CF-205 K Chest Freezer) at minus 20°C until testing (see Figure 3.2).

A study done by Roe *et al.*, (1988) showed that canine bones can be stored at minus 20°C for up to 32 weeks without showing any significant structural changes<sup>34</sup>.



**Figure 3.2:** Individual storage of specimens. Specimens were individually wrapped in cotton towels, soaked in lactated Ringers solution, and sealed in marked plastic bags.

### 3.4.5 Anatomical components of the specimens

The final spine bony segment consisted of intact spinal segments L5 to L7 and the sacrum.



### 3.4.6 Assignment of the specimens

The spinal specimens were randomly assigned into two groups (nine per group) as shown in Table 3.2.

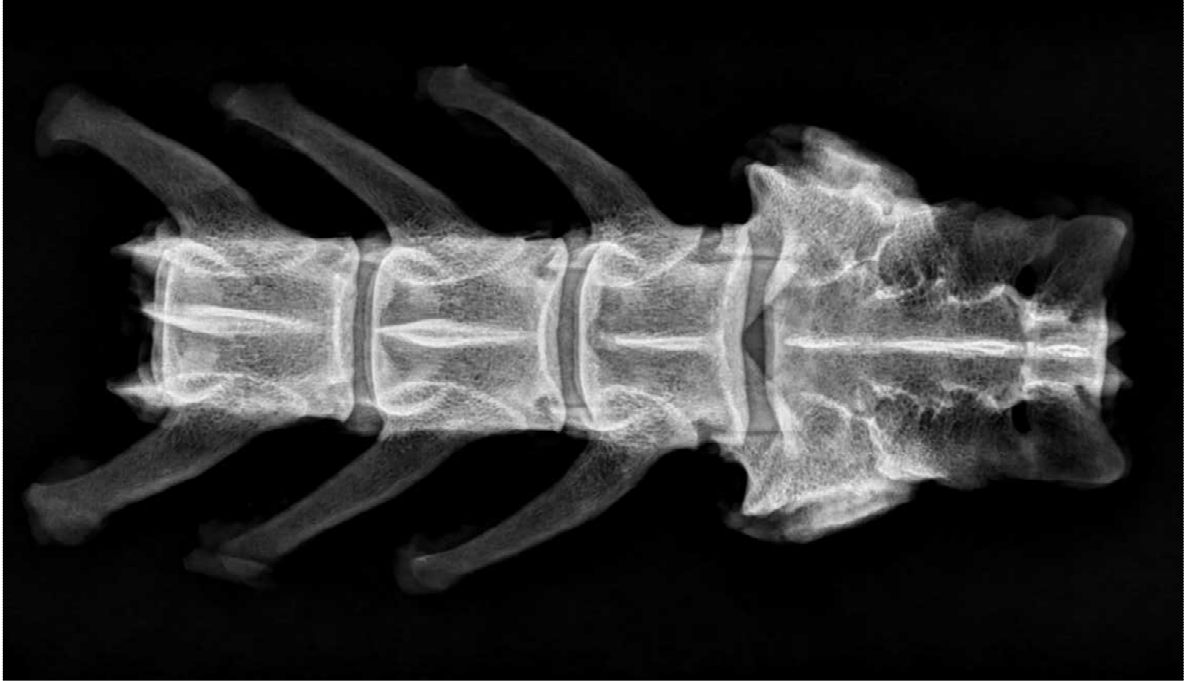
PMMA (n=9)	SOP (n=9)
s1	s2
s3	s4
s6	s5
s8	s7
s9	s10
s11	s12
s13	s14
s16	s15
s17	s18

**Table 3.2:** Assignment of specimens to the two fixation methods

## 3.5 Radiographs and Photographs

Orthogonal radiographs (lateral and dorso-ventral views) were made of all the spinal specimens to assess the spinal column for any sign of degenerative lumbosacral pathology (see Figure 3.3 & 3.4) by using a Toshiba Rotanode EVA-HF 325/525 high frequency radiography X-ray system.

Exposure factors used were 48 kV and 8 mAs. Radiographs were digitally developed and stored in DICOM format at the Fourways Veterinary Hospital. Initial screening was done by the author and any bone specimen with obvious lumbosacral pathology was discarded.



**Figure 3.3:** Dorso-ventral radiographic view of the spinal segment L5-S3 showing no degenerative changes. The cranial end (L5) is to the left and the sacrum is to the right in this image.



**Figure 3.4:** Left lateral radiographic view of the spinal segment L5-S3, showing no degenerative changes. The cranial end (L5) is to the left and the sacrum is to the right in this image.

No people were present inside the radiography room during the radiographic procedures and therefore no person was exposed to radiation. All the digital photographs were taken using a Canon digital camera. (Canon® *IXUS 245 HS 16.1 Megapixels*)

### 3.6 Defect condition (Osteotomy of L7)

The interspinal ligament between L7 and S1 was sharply resected. In all the spines, a complete spinal fracture-luxation model was produced at the L7-S1 junction by transecting all the ligamentous structures connecting the two vertebrae. An oblique osteotomy through the caudal aspect of the vertebral body of L7 was made using an oscillating saw (*Stryker 2296-34 Sagittal Saw*; Stryker Corporation, Kalamazoo, Michigan) to simulate a fracture in this area, as typically seen in clinical cases (see Figure 3.5). The reproduction of similar osteotomies – although not identical, due to individual variation between specimens – of the 7<sup>th</sup> lumbar vertebral body in a consistent fashion *in vitro* was readily achieved, making comparison of the specimens meaningful.



**Figure 3.5:** Lateral radiograph of the caudal lumbar spine, illustrating a typical lumbosacral fracture–luxation. A small oblique fracture of the caudal body of L7, with typical cranio-ventral displacement of the caudal fragment and sacrum is visible.

### 3.7 Fixation techniques (Osteotomy fixation)

Surgical stabilisation using pin-PMMA or SOP locking plates was done after the osteotomy. To allow for more accurate comparison between the different specimens in the two groups, the diameter of the positive profile half threaded pin, the volume of the PMMA used, and the size of the SOP bone plate was standardised throughout. A mounted skeleton was available at all times for reference purposes.

#### 3.7.1 pin-PMMA

Nine spinal fracture-luxation models were randomly selected for fixation using four positive profile end-threaded pins anchored in the vertebral bodies of L7 and S1. These pins were bonded dorsally with PMMA (Group 1) as described by Weh *et al.*, (2007)<sup>46</sup>.

The fracture-luxation was reduced and the L7–S1 facet joint was aligned and used as a guideline to assess the accuracy of reduction.

Stabilisation was performed using 3.4 mm positive profile end-threaded pins (Röth Medical Components (Pty) Ltd, Cape Town, South Africa) which were placed in the pedicle/body of L7 and in the body of S1. All pin holes were predrilled with the appropriate size drill bit.

Méheust *et al.*, (2000) used a pedicle screw fixation to stabilise the lumbosacral area. He determined that the ideal point of entry in L7 is located at the small bony crest, at the base of the cranial articulation facet, placed parallel to the sagittal plane. For S1, the ideal point of entry is at the sloped plane just caudal to the cranial articulation facet, with the screw placed parallel to the vertebral end plate of the sacrum<sup>25</sup>.

The pins in L7 were inserted caudal to the base of the *Processus articularis cranialis* and directed cranio-ventro-medially (see Figure 2.7), exiting at the cranio-ventral vertebral body.

This positioning allowed maximum purchase of the pins in the pedicle and vertebral body and, at the same time, avoids the area where these lumbosacral fracture–luxations are typically seen in the clinical situation – usually a small oblique fracture of the caudal body of L7<sup>41</sup>.

For the lumbar pins, only the outer cortex of the pedicle was penetrated with a drill initially. A 1.0 mm Kirschner wire was then used to probe the cancellous bone of the pedicle to ensure that the pins would not compromise the medial pedicle cortex as described by Weh *et al.*, (2007)<sup>46</sup>. Following the trajectory of the previously inserted Kirschner wire, the drill was then advanced until the ventral cortex of the vertebral body was penetrated.

A depth gauge was used to measure the hole depth and the same distance was measured along the threaded portion of the pin to determine proper depth of pin insertion.

The caudal pins which were placed into S1 were inserted just caudal and lateral to the caudal articular process of L7 and directed caudo-ventro-laterally through the sacrum.

All the pins were placed to exit 2-3 mm ventral to the vertebral body (see Figure 3.6).



**Figure 3.6:** Photo of the ventral view of pin placement. Pins were placed to exit 2-3 mm ventral to the vertebral body (arrows). The cranial end (L5) is to the left and sacrum is to the right in this image.

Notches were cut into the fixation pins using a pin cutter, and the pins were then bent to achieve maximum overlap of the caudal, lateral, right, and left pins. The pins were cut just below the level of the dorsal spinous processes' dorsal edge (see Figure 3.7).



**Figure 3.7:** Photo of a right lateral view of the specimen after placement of the pins for the pin-PMMA method. The pins were bent to achieve maximum overlap of caudal, lateral, right, and left pins at a level just below the dorsal spinous processes' dorsal edge. PMMA was applied dorsally to bond all the pins, articular facets, and dorsal spinous processes of L7 and S1. The cranial end (L5) is to the left and the sacrum to the right in this image.

Due to the relatively high cost of medical grade PMMA, dental grade non-sterile PMMA powder and liquid monomer (Excel, Wright Health Group, UK) was used in this study.

Haas *et al.*, (1975) found that the mechanical properties of medical grade PMMA approximated those of dental grade non-sterile PMMA<sup>17</sup>. Forty grams of non-sterile PMMA powder and liquid monomer (Excel, Wright Health Group, UK) was measured and used as suggested for dogs weighing more than 15 kilograms<sup>8</sup>.

The liquid polymer was added to the polymer powder as per manufacturer's instructions and mixed for approximately four minutes, until a doughy consistency was reached and the mixture no longer adhered to the surgical gloves. The PMMA was moulded in a cylindrical

mass and applied dorsally to bond all the pins, articular facets, and dorsal spinous processes of L7 and S1 (see Figure 3.8 & 3.9). The hardening PMMA was lavaged with cooled saline for about five minutes after placement to avoid any potential negative effects on the bone caused by the heat from the exothermic reaction. The PMMA column was then allowed to harden for more than 24 hours before testing.



**Figure 3.8:** Photo of the dorsal view of the pin-PMMA placement. The cranial end (L5) is to the right and the sacrum is to the left in this image.



**Figure 3.9:** Photo of the lateral view of the pin-PMMA placement. The cranial end (L5) is to the left and the sacrum is to the right in this image.

Ventro-dorsal and lateral radiographs were taken to ensure correct fracture-luxation reduction and pin placement. (Figure 3.10 & Figure 3.11)



**Figure 3.10:** Lateral radiographic view of the pin-PMMA placement. The cranial end (L5) is to the left and the sacrum is to the right in this image.



**Figure 3.11:** Dorso-ventral radiographic view of the pin-PMMA placement in L7 and S1. The cranial end (L5) is to the left and the sacrum is to the right in this image.



### 3.7.2 SOP Plating

Nine spinal fracture-luxation models were randomly selected for fixation using 3.5 mm SOP Locking Plates (Orthomed UK Ltd, Halifax, UK) with 3.5 mm cortical screws (Orthomed UK Ltd, Halifax, UK), anchored in the L6, L7, S1, and S2 vertebral bodies. Two screws were placed in L6 (cranially and caudally), one screw was placed in the cranial part of L7, and another screw was placed in S1. One screw was also placed into S2 after the SOP screw hole was pre-drilled into S2 with a 2.5 mm Kirschner pin. This was repeated on the contralateral side to achieve bilateral dorso-lateral positioning of the implants.

The manufacturer recommends using two bilaterally placed SOP plates, applied to the dorsal-lateral aspect of the spinal column. Recently, it was recommended that SOP locking plates, used in a bilateral configuration, be twisted caudally to also engage the shaft of the ilium with a recommended four screws – three screws at a minimum – cranial to the fracture and four screws – three screws at a minimum – caudal to the fracture<sup>22</sup>. The configuration used in the present study was selected based on the configuration used at Onderstepoort Veterinary Academic Hospital before the above recommendations were available.

The 3.5 mm SOP Interlocking plates were pre-operatively contoured using specific bending irons (Orthomed UK Ltd, Halifax, UK). Bending tees were placed inside the screw holes to prevent deformation of the screw holes during bending (see Figure 2.4)

The pre-contoured SOP plates were positioned on the dorso-lateral aspect of the vertebrae, at the level of the base of the transverse processes of the lumbar vertebrae.

Because torsion can be applied to the internodes, fine individual adjustments were made as needed before drilling into the vertebrae.

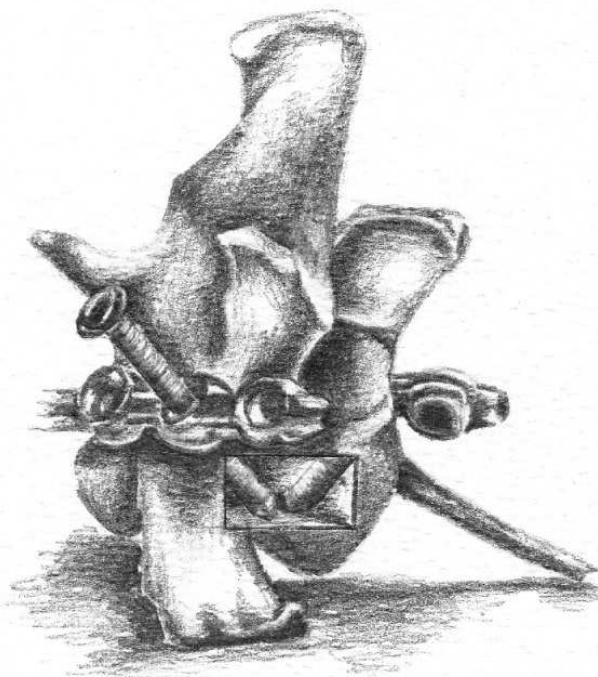
During preparation, the specimens were kept moist by regular spraying with lactated Ringers solution (Sabax Ringers Lactate®, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg).

Hypodermic 22 gauge needles were used to clearly identify the location of the intervertebral discs.

For the lumbar screws, only the outer cortex of the pedicle was penetrated with a drill initially. A 1.2 mm Kirschner wire was then used to hand probe the cancellous bone of the pedicle, in order to ensure that the screws did not compromise the medial pedicle cortex as described by Weh *et al.*, (2007)<sup>46</sup>. Following the trajectory of the previously inserted Kirschner wire, the drill was then advanced until the ventral cortex of the vertebral body was penetrated.

A depth gauge was used to measure the depth of the bony tunnel and the correct length of 3.5 mm cortical screw was selected.

The screws were directed at an angle of approximately 60° from the mid-sagittal plane into the vertebral body of L6<sup>43; 45</sup>. (See Figure 3.12)



**Figure 3.12:** Screw placements in the L6 vertebral body. Bilateral SOP plates are applied on the dorso-lateral aspect of the spine, with the screws directed at an angle of 60° (range 55°-65°) from the mid-sagittal plane into the lumbar vertebral bodies L1 to L6. (Source: L. Liebenberg; Fourways Veterinary Hospital, Bryanston, South Africa)

At the level of L7, the pedicle is wide enough to allow for the implantation of pins or screws and the technical difficulty caused by the iliac wings is avoided<sup>47</sup>. The entry point used was located at the base of L7's cranial articular process, and the orientation of the screw was parallel to the sagittal plane (see Figure 3.13). The entry point used for S1 was located a few millimetres behind its cranial articular surface, with the screw directed about 5° relative to the sagittal plane (see Figure 3.13). Bilateral screws were also placed into S2 after the SOP screw hole was drilled into S2 with a 2.5 mm Kirschner pin, instead of a drill bit.

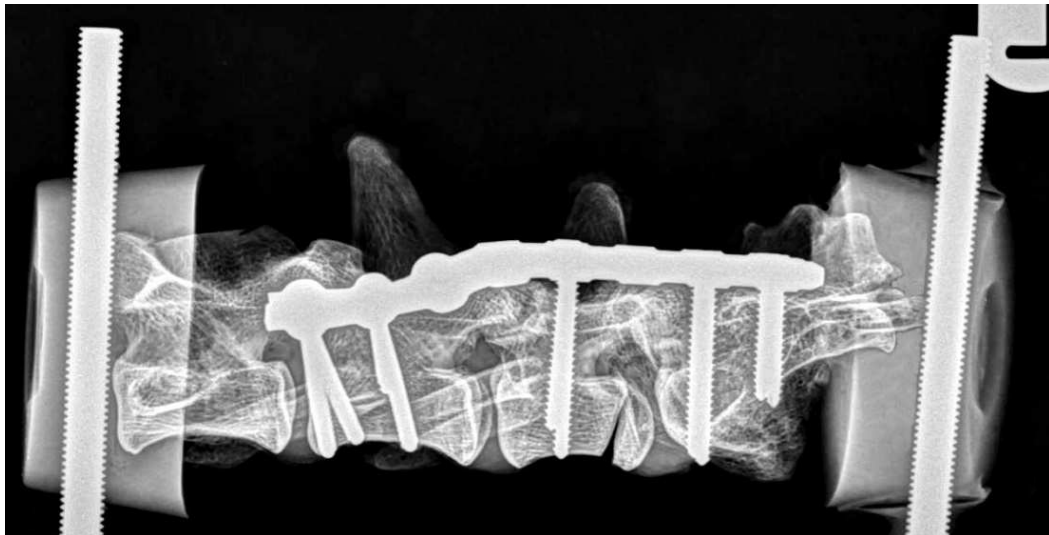


**Figure 3.13:** Photo of the dorsal view of the SOP placement. Stabilisation was performed using 3.5 mm SOP Locking plates (Orthomed UK Ltd, Halifax, UK) bilaterally, with 3.5 mm cortical screws were anchored in the L6, L7, S1 and S2 vertebral bodies. The cranial end (L5) is to the left and the sacrum is to the right in this image.



**Figure 3.14:** Photo of the lateral view of the SOP placement. The cranial end (L5) is to the left and the sacrum is to the right in this image. Note how the SOP plates are bent along the contours of the vertebrae to obtain an optimal fit.

Accurate pre-contouring was still important because, unlike DCP plates where the screw angle can be adjusted to some degree, the nodes of the locking plate design determine the direction of screw placement (see Figure 3.14). Ventro-dorsal and lateral radiographs were made after complete fracture-luxation stabilisation to ensure correct fracture-luxation reduction and proper implant placement (see Figure 3.15 & Figure 3.16).



**Figure 3.15:** Lateral radiographic view of the SOP placement taken to ensure correct reduction of the fracture luxation and placement of the SOP plate and screws. The cranial end (L5) is to the left and the sacrum is to the right.



**Figure 3.16:** Dorso-ventral radiographic view of the SOP placement taken to ensure correct reduction of the fracture luxation and placement of the SOP plate and screws. The cranial end (L5) is to the left and the sacrum to the right.

### 3.8 Testing

As with many tests, the result, behaviour, and response of the test specimens will most likely be influenced by the test conditions, such as loading, constraints, and environmental factors. For this reason, it is of great importance to ensure that the test conditions are carefully considered and documented in order to ensure correct interpretation of the test results and comparison with the results obtained from other tests.

From the current literature, it does not seem that there is a standard testing procedure to be followed for the testing of canine lumbosacral cadaver specimens. Early *et al.*, (2013) stated that conflicting results were reported by Meij *et al.*, (2007) and Smoulders *et al.*, (2012)<sup>14; 26; 38</sup>. Early *et al.*, (2013) then tried to evaluate this by using another test setup, a cantilever beam, instead of the 4-point bending setup as used in the other studies<sup>14</sup>.

In the testing of previously reported bony spine segments, there were variations in crosshead speeds, ranges of motion, and loading configurations. The manner in which the spine is loaded will also affect the response of the spine segment. If loading on the spine differs between two types of test rigs, then comparisons can only be made between the tests done on the same machine. However, comparisons cannot be made by using results obtained using a different test rig. It would therefore be advantageous if a standardised test could be established in order to make the comparison of different tests possible. This could then make it possible to compare the results from tests performed on similar spine segments with regard to aspects such as the range of motion of an intact spine and the stabilisation with different fixation techniques. Recommendations for the standardisation of testing procedures for the different spinal implants that are currently in use have been suggested by Wilke *et al.*, (1998)<sup>50</sup>.

Meij *et al.*, (2007) tested a similar canine lumbosacral segment: L5-L7, the sacrum, and the first caudal vertebrae<sup>26</sup>. The specimens were subjected to a bending moment in the sagittal plane using a 4-point bending device. The vertical tension and compression load was

applied by using a hydraulic materials testing machine, resulting in the flexion and extension of the spine specimen. The 4-point bending device then converted the vertical movement of the crosshead of the machine to a rotation about a fixed point on either side of the spine. The vertical displacement of the crosshead of the materials testing machine was equivalent to the angle of bending in the 4-point bending test. Bending angles and moments were determined from displacements and loads measured by the materials testing machine. The materials testing machine was displacement controlled with a resulting constant angular velocity of 0.3°/s (implying a crosshead speed of 0.3 mm/s). The load was applied until 3 Nm was reached. The bending started with flexion until 3 Nm, after which the loading direction was reversed until the spine was loaded in extension to -3 Nm. Each spine segment had three series consisting of recordings of five loading cycles each.

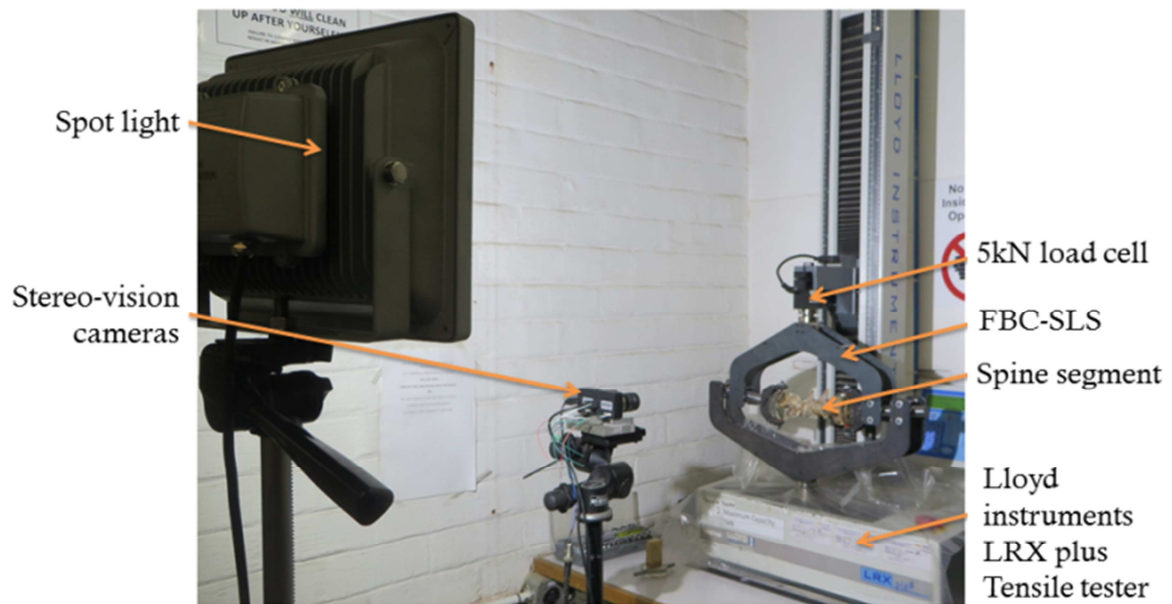
A potential complication of the 4-point bending device used by Meij *et al.*, (2007) is the development of forces along the cranio-caudal axis of the spine <sup>26</sup>. This will imply that the loading of the spine will change from being in pure bending to a combined loading of bending and cranio-caudal loads.

The testing procedure that was followed for the tests performed in this study was based on recommendations of Wilke *et al.*, (1998) <sup>50</sup>.

### **3.9 Test equipment**

Figure 3.17 shows the experimental setup that was used during the biomechanical testing of the spinal column specimens. A specially designed Free Bending Canine Spinal Loading Simulator (FBC-SLS) was connected to the Lloyd instruments LRX plus tensile tester (Ametek Inc., Berwyn, PA, USA) via the top and bottom jaws. A 5kN load cell was placed between the tensile tester and the top jaw. The tensile tester actuated the top jaw, thereby flexing or extending the spine segment. The spotlight was used to ensure sufficient lighting of the specimen as well as the FBC-SLS. The displacement of the tensile tester and the

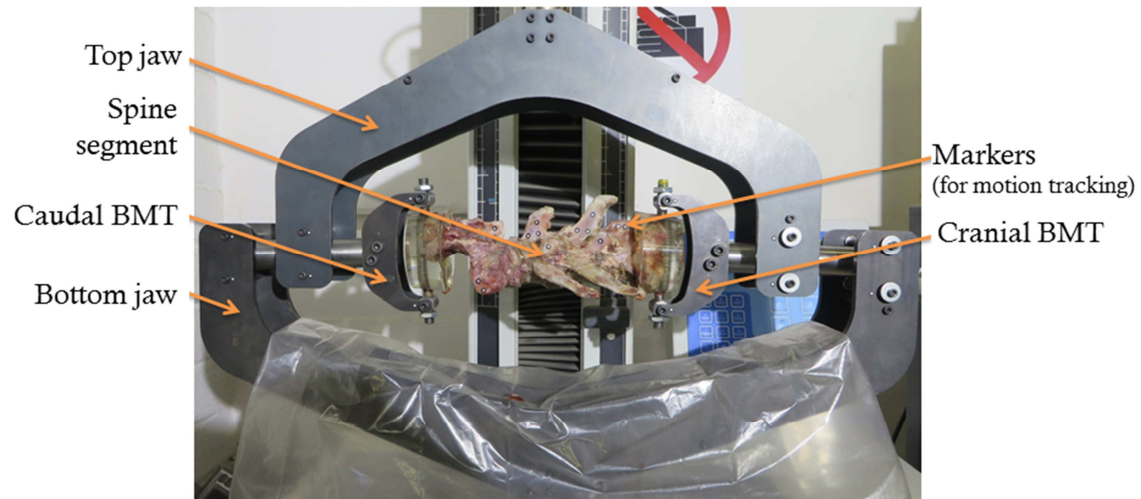
force in the 5kN load cell were recorded at a sampling frequency of 100Hz, with the motion system recording data at a sampling frequency of 7.5Hz.



**Figure 3.17:** Experimental setup used during the biomechanical testing of the spine specimens. The components used are annotated on the photo.

### 3.9.1 Spinal loading simulator

The spine segment was loaded using the FBC-SLS shown in Figure 3.18. The FBC-SLS aimed to load the spine segment in flexion and in extension as realistically as possible. To achieve this, the FBC-SLS loaded the spine segment with a pure bending moment at the cranial and caudal ends of the bony spine specimen. This occurred in the sagittal plane, without constrain on the cranio-caudal axis and the bending moment allowed translation along the cranio-caudal axis and rotation about this axis. The spine segment was connected to the FBC-SLS by potting L5 and S1 in epoxy (*DomeX<sup>®</sup> Epoxy and Hardener; Hi Tech Polymers, Irene, Pretoria*) to a threaded rod. The threaded rods were then connected to the caudal and cranial Bending Moment Transmitter (BMT). The caudal end of the specimen was on the left side of the FBC-SLS and the cranial end on the right side.



**Figure 3.18:** The Free Bending Canine Spinal Loading Simulator (FBC-SLS). The components used are annotated on the photo. The cranial end (L5) is to the right and the sacrum is to the left in this image.

### 3.9.2 Motion measurement system

The motion of each vertebra in the bony spine segment was measured using a motion measurement system which made use of stereo-vision cameras (Point Grey's Chameleon 2 camera (Point Grey Research Inc., Richmond, BC, Canada)). From the motion data of the individual vertebra, the different joint angles could be calculated. In many of the specimens it was not possible to place markers on L5, as this vertebra was potted in epoxy in order to connect it to the FBC-SLS. It was therefore necessary to track the motion of the cranial BMT and use the motion of the cranial BMT as the motion of L5. This implies that it was assumed that the connection between L5 and the cranial BMT through the epoxy and rod was a rigid connection. This was deemed a valid assumption as a very good bond was obtained between L5 and the epoxy.

### 3.10 Test conditions

This section reports the conditions during testing.



### **Temperature**

Higher temperatures, such as body temperature, generally accelerate the cellular autolytic processes, shortening the possible test duration as it compromises the specimen's biomechanical properties<sup>50</sup>. Temperature of the testing environment in the present study was controlled by setting the air conditioning unit to 25°C.

### **Test duration**

Wilke *et al.*, (1998) advised that biomechanical tests on bone should not be performed over more than 20 hours of room temperature exposure, as the properties of the specimens begin to change beyond this point<sup>50</sup>. All attempts were made to ensure that exposure to room temperatures during the testing procedures was limited as far as was practically possible.

### **Moisture condition**

During preparation and testing, the bone specimens were kept moist by regularly spraying with lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg).

### **Loading rate**

The crosshead speed of the Lloyd actuator was set at 90 mm/min for all tests. For the FBC-SLS this implied a deformation rate of the specimen of 1.455%/s, calculated from the displacement-angle sensitivity of the Free Bending Canine Spine (FBCS) test rig. This deformation rate falls within the recommended range of 0.5%/s to 5.0%/s<sup>50</sup>.

## **3.11 Specimen preparation**

Individual specimens were prepared for mounting to the FBCS-SLS. This was done by potting the cranial and caudal ends of the segment in an epoxy (*DomeX*<sup>®</sup> *Epoxy and Hardener*; *Hi Tech Polymers, Irene, Pretoria*). The containers for the potting process were made of 60 millilitre (8 cm diameter and 5 cm high) plastic cups.

Each cup contained an 8 mm threaded rod that penetrated it from side to side. The threaded rods were used to connect the bony specimen to the FBCS-SLS.

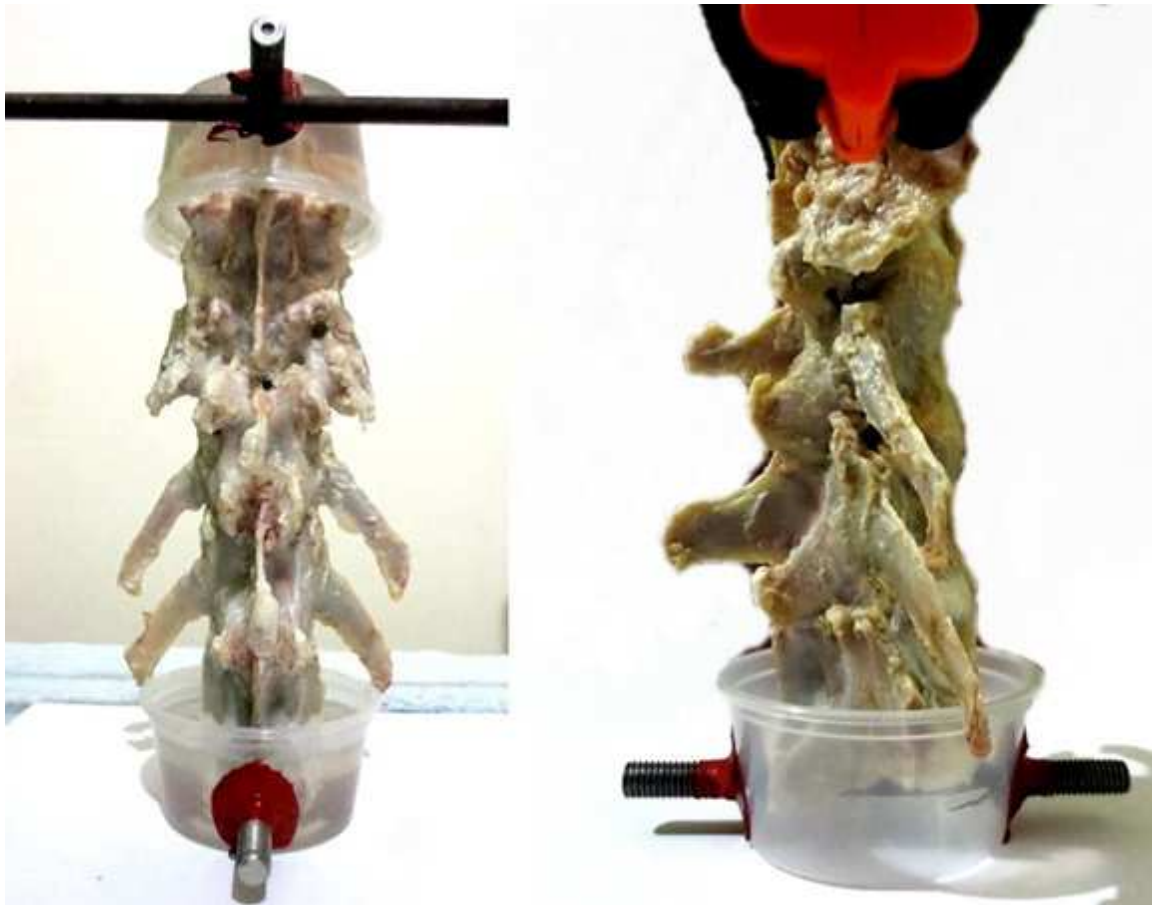
The potting procedure was as follows:

- Each bony specimen was removed from the freezer and thawed for 30 min in lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg) at a room temperature of 25 °C.
- During the thawing of the bony specimen, the epoxy (*DomeX*<sup>®</sup> *Epoxy and Hardener; Hi Tech Polymers, Irene, Pretoria*) was prepared.
- After the specimen had thawed for 30 min, the sacral end of the specimen was dried using paper towels and then potted into the epoxy.
- Care was taken during the potting process to ensure that the spinal segment was aligned with the metal rod on both the cranial and sacral ends of the bony specimen and that it was positioned exactly in the median plane (see Figure 3.19).



**Figure 3.19:** Dorsal and lateral views of potting and alignment by visual means of the sacral end of each bony specimen.

- The sacral end was allowed to set for 90 minutes at room temperature (25°C), which was sufficient time for the epoxy to cure to such a degree that the specimen could be flipped so that preparations could be made to the cranial end.
- The cranial end (L5) was dried using paper towels and potted in the epoxy.
- Care was taken during the potting process to ensure that the spinal segment was perfectly aligned with the rod on both the cranial and sacral ends and that it was positioned exactly in the median plane (see Figure 3.20)
- The cranial end was also allowed to set for 90 minutes.
- Each specimen was then stored as described in 3.4.4 Storage of specimens until testing.



**Figure 3.20:** Dorsal and lateral view of potting and alignment by visual means of cranial end of the specimen.

During potting of each bony specimen, the specimen was kept moist by regularly spraying it with lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg). Total time that the specimen was at room temperature during potting was approximately 210 minutes.

Before testing, the prepared bony spinal segments were removed from the freestanding domestic freezer (*LG*<sup>®</sup> *CF-205 K Chest Freezer*) at minus 20°C and placed in a bar fridge (*National NR-A7M2HBar Fridge*) for 24 hours at 4°C to thaw. Nine specimens were tested per day and the testing period extended over two days. On the day of testing, nine specimens were packed in a cooler box (*Coleman*<sup>®</sup> *16 Quart Excursion*<sup>®</sup> *Cooler*) and were transported to the Department of Mechanical and Aeronautical Engineering, University of Pretoria. On arrival, the specimens were immediately transferred to a bar fridge (*National*

*NR-A7M2HBar Fridge*) where they remained for the period of testing at 4°C. Before testing each specimen, it was first removed from the plastic bag and placed in a bath with lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg) at room temperature ( $\pm 25^{\circ}\text{C}$ ) for 30 minutes.

### **3.12 Preparation of the testing equipment**

The following preparations were made with respect to the test equipment:

- Care was taken to ensure that all Allan head bolts on the bottom jaw of the FBC-SLS were properly tightened.
- It was ensured that the 5kN load cell was properly fastened to Lloyd 5kN tensile tester.
- The top jaw on Lloyd 5kN tensile tester was aligned by placing a steel ruler between the top support of the bottom jaw and the bottom supports of the top jaw (see Figure 3.21).
- All the bolts were tightened and locked.



**Figure 3.21:** Alignment of top and bottom jaws of the FBC-SLS. A steel ruler was used to align the top and bottom jaws of the FBC-SLS.

### 3.13 Loading of each bony specimen

The values presented in Table 3.3 were used in a triangular wave displacement signal with the specific amplitudes in tension (results in flexion of the specimen) and compression (results in extension of the specimen) created and used as input to the FBC-SLS. In addition to the displacement controlled input signals, a force controlled input signal was also created. This signal was again of a triangular wave shape, with the same force amplitude in extension and compression. The input signals consisted of five cycles, with each cycle consisting of full extension and flexion. All the specimens were first loaded in flexion.

Load range	Extension (-)	Flexion(+) [mm]	Control
3Nm	100 [N]	100 [N]	Force
lr0	6.5 [mm]	14 [mm]	Displacement
lr1	8.75 [mm]	18.5 [mm]	Displacement
lr1p5	8.75 [mm]	30 [mm]	Displacement
lr2	12.5 [mm]	30 [mm]	Displacement

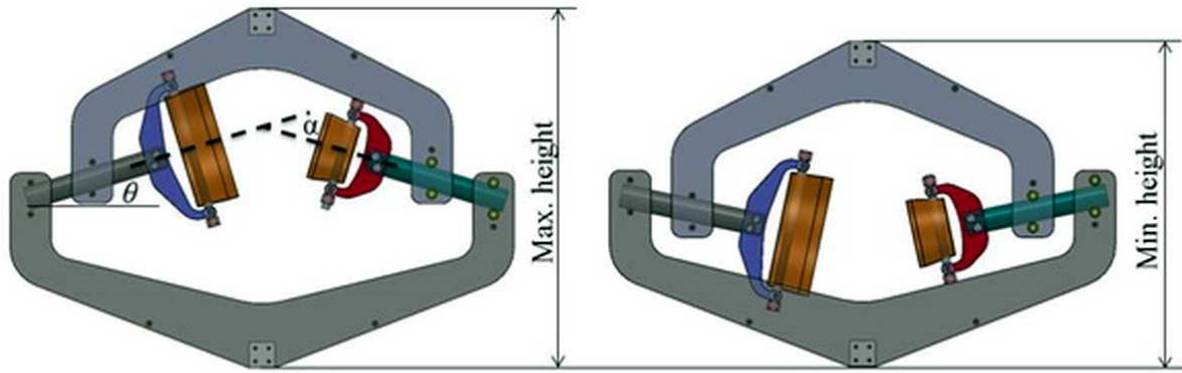
**Table 3.3:** Load ranges used for testing fixated spine specimens

### 3.14 Testing procedure

The order of testing the spinal column segments was randomly assigned. The specimen was removed from the bath of lactated Ringers solution (*Sabax Ringers Lactate*<sup>®</sup>, Adcock Ingram Critical Care (Pty) Ltd, Johannesburg) at room temperature ( $\pm 25^{\circ}\text{C}$ ) and mounted to the FBC-SLS. The specimens were mounted to the FBC-SLS via the epoxy-rod connectors on the two ends of the segment in which the spine segment was potted. Markers were placed on the specimen for motion tracking. After the specimen had been mounted and the markers were placed, the FBC-SLS and specimen were moved to the starting position. The starting position was set as follows:

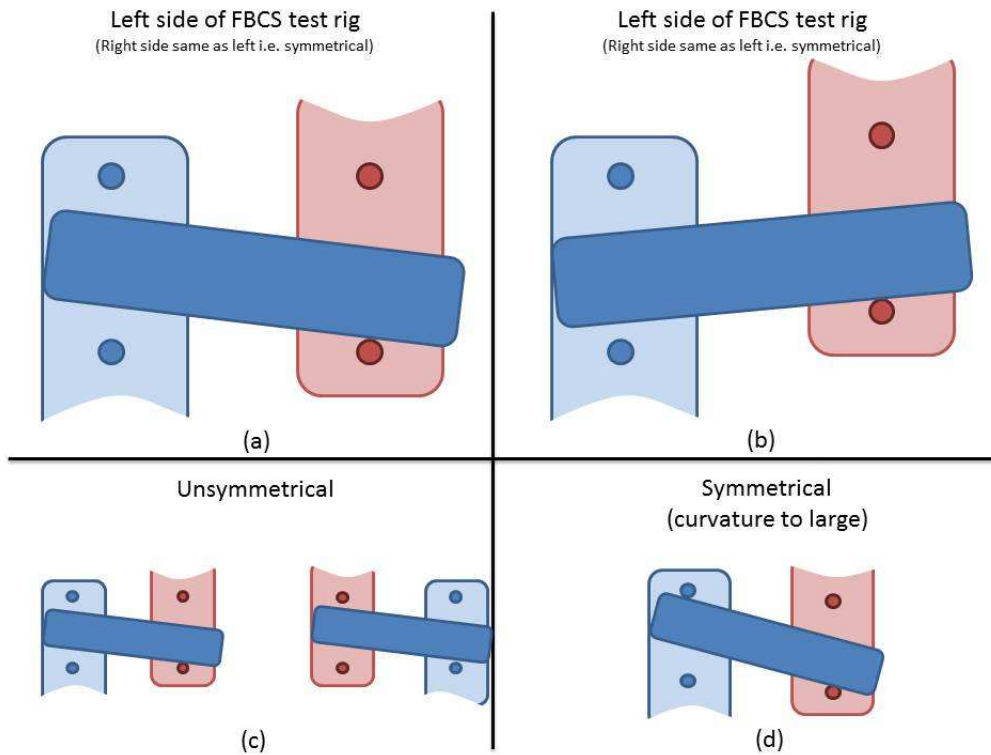
- The 5kN load cell was zeroed, with only the top jaw connected to the load cell.
- The top jaw was moved so that the BMT's only made contact with the bottom support of the top jaw (see the bottom red dot in Figure 3.22).
- The load on the load cell was noted.
- The top jaw was moved down until the load cell's measurement just started to change. This implied that the BMT was starting to make contact with the bottom support of the bottom jaw (see the bottom blue dot in Figure 3.22).
- The load cell and displacement on the Lloyd 5kN tensile tester were zeroed.

Once the FBC-SLS and specimen were at the starting position, the specimen was loaded with one of the load ranges specified in Table 3.3.



Flexion

Extension



Left side of FBCS test rig  
(Right side same as left i.e. symmetrical)

Left side of FBCS test rig  
(Right side same as left i.e. symmetrical)

(a)

(b)

Unsymmetrical

Symmetrical  
(curvature to large)

(c)

(d)

**Figure 3.22:** Position at which the weight of the BMT's and bony spine segment is measured  
Blue dots are associated with the bottom jaw and red dots are associated with the top jaw.



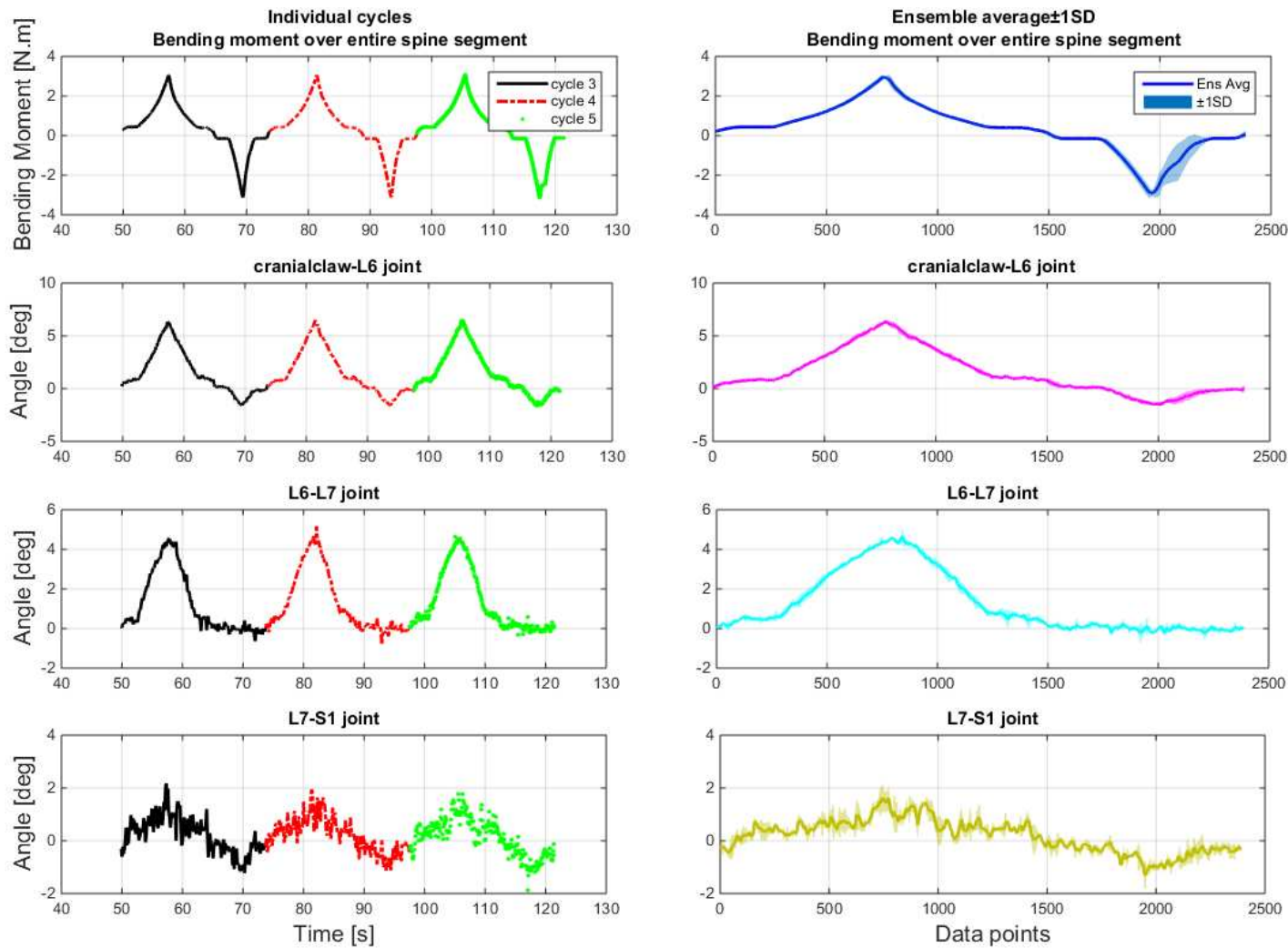
## 3.15 Data and statistical analysis

### 3.15.1 Biomechanical parameters

The recorded motion data was processed in order to obtain the angle of each joint of the bony spinal segment. The force data was used to calculate the bending moment that was imposed over the spine segment. The first two cycles of the data were excluded as these only conditioned the bony specimen, with the specimen being fully conditioned by the third cycle as described by Wilke *et al.*, (1998)<sup>50</sup>.

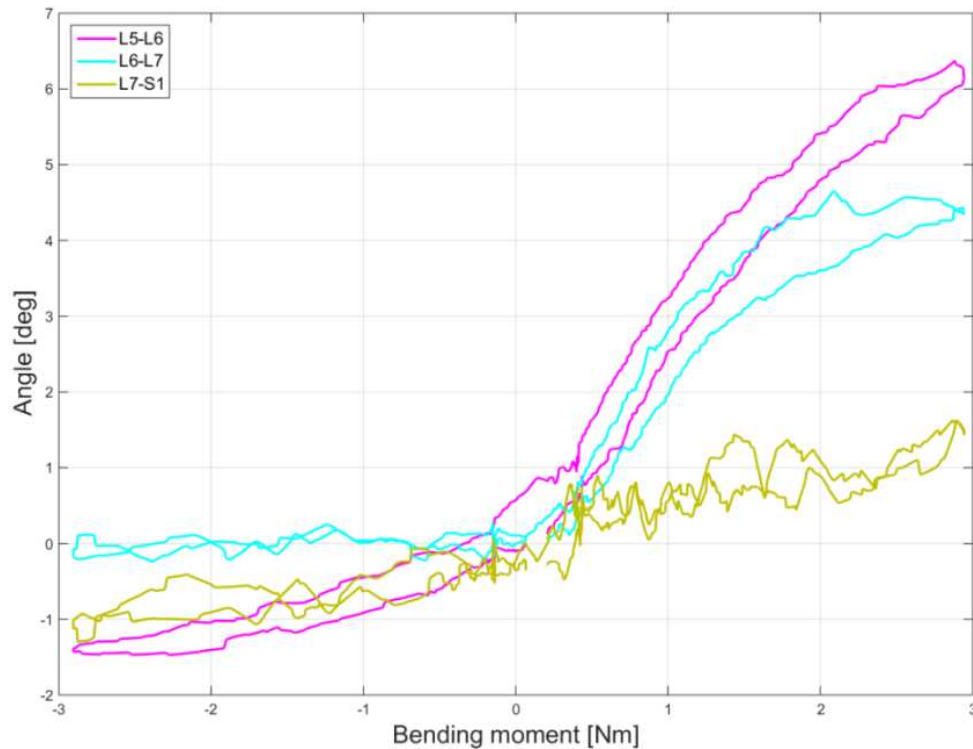
The third, fourth, and fifth cycle of the angular displacement of each joint (joint L5-L6, L6-L7 and L7-S1) of the bony spine specimen and the bending moment that was acting over the specimen was extracted and used to calculate the ensemble average and standard deviation of the joint angle and bending moment. The ensemble average and standard deviations were calculated for the three cycles of each specimen and are referred to as the ensemble average and standard deviation of the cycles (Ens. Avg. and SD (cycles)).

The process described above is shown graphically in Figure 3.23.



**Figure 3.23:** Graphs of the cycles of the angle and bending moment of each joint with the associated ensemble average and standard deviation (Specimen 6 fixated, Load range: 3Nm)

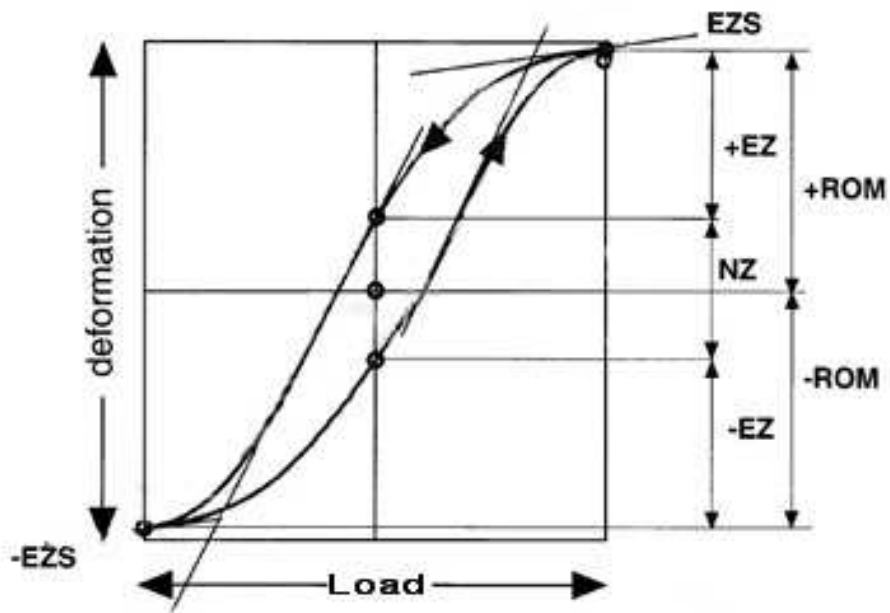
By using the ensemble average (cycles), bending moment and joint angles, the bending moment-angle characteristic of each joint was obtained as depicted in Figure 3.24.



**Figure 3.24:** Graph depicting the bending moment-angle (Ens. Avg. (cycles)) characteristics of the three joints (Specimen 6, pin-PMMA, 3Nm).

The following biomechanical parameters were extracted from the bending moment-angle characteristic as specified in Wilke *et al.*, (1998) and used by Meij *et al.*, 2007 (see Figure 3.25) <sup>26; 50</sup>.

These were defined for each of the two test directions: flexion (+) and extension (-):



**Figure 3.25:** Graph depicting the load-displacement curve along which all specimens were evaluated.  
(Printed with permission from Wilke H J, et al., (1998), "Testing criteria for spinal implants: recommendations for the standardisation of in vitro stability testing of spinal implants", *European Spine Journal*, 7:148-154). (Appendix 4)

**Neutral zone (NZ)** – The difference in angulation between the two phases of motion at zero loading. This was a measurement of the laxity of the bony spinal specimen, describing the range over which the specimen moved when free of any applied loading.

**Elastic zone (EZ)** – The deformation measured from the end of the neutral zone to the point of maximal loading.

**Range of motion (RoM)** – A description of the sum of the neutral zone and the elastic zone in one direction of motion (i.e. in flexion or extension).

**Elastic zone stiffness (EZS)** – The stiffness characterising the elastic deformation of the bony specimen or construct.

### 3.15.2 Statistical techniques used

Descriptive statistics such as means and standard deviations, and graphical representations such as box plots were used to get a profile of the different or various measurements.

The mean and standard deviation of the biomechanical parameters were calculated for each joint over all the bony spine specimens in the two fixation groups. As the sample sizes were small, non-parametric statistical tests were performed. To compare the means of the biomechanical parameters of the two fixation techniques, the Wilcoxon rank sum test was performed and used. Comparisons of the means of more than two repeated measures (measurements across the L5-L6, L6-L7 and L7-S1 vertebral bodies) were performed using the Friedman test for repeated measures.

Non-parametric statistical techniques used to compare the means were based on ranks and were developed specifically for small samples.

### 3.15.3 Data excluded from analysis and statistics

With regarding to the pin-PMMA technique, the data of two fixated specimens were not included in the analysis because the one's motion data was not usable and the other one had a large amount of PMMA in the epoxy dorsal to S1. Out of the remaining seven specimens, the data of joint L7-S1 of specimen No. 9 was also excluded because the motion data was not usable. All the data for the nine specimens in the SOP fixated group was included in the analysis.

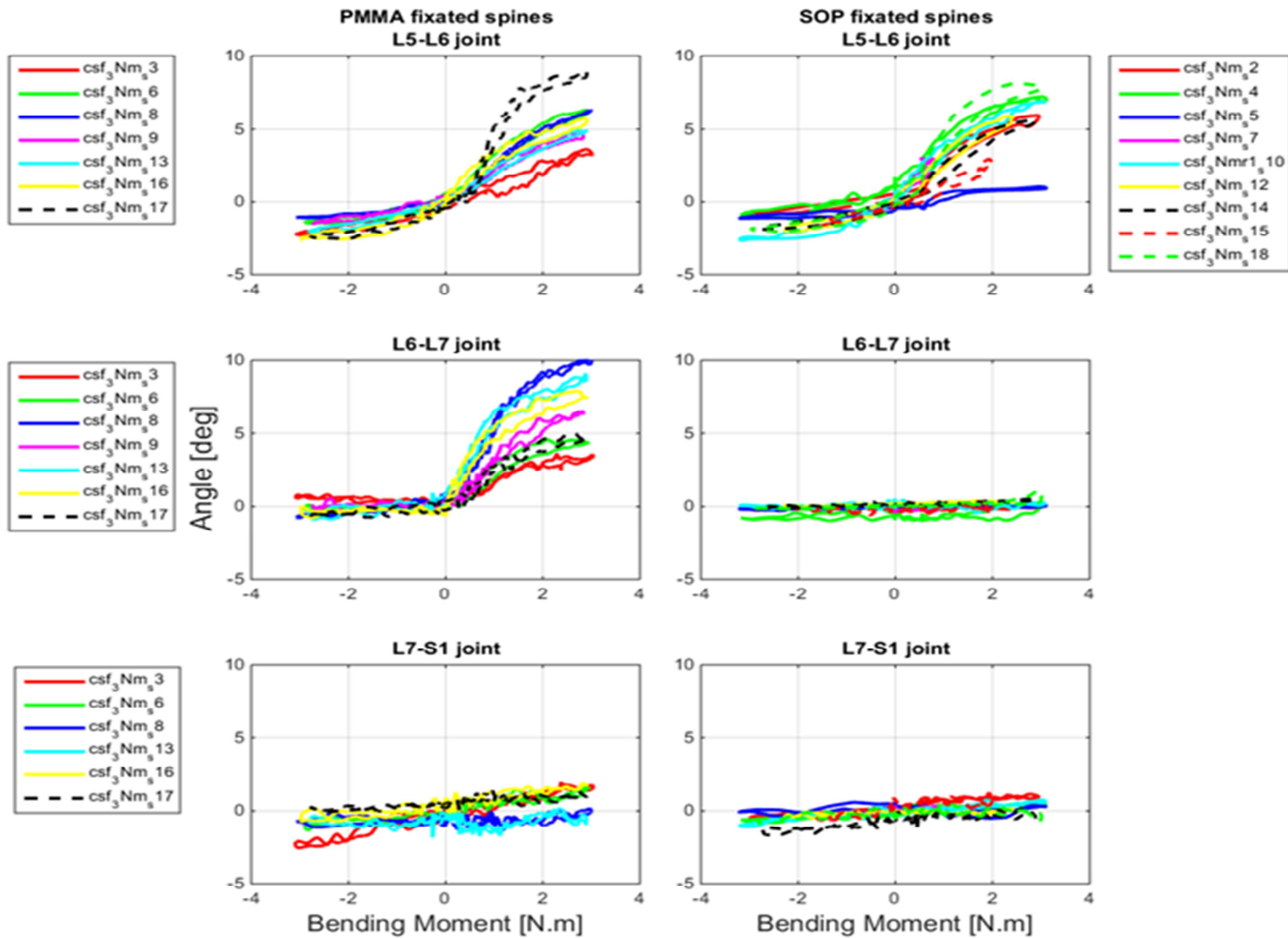
Specimen	Data excluded (specimen/joint)	Reason
s1, s11	Bone specimen	Significant amount of PMMA in epoxy at the caudal end and motion data not usable.
s9	Joint L7-S1	Motion tracking not able to track points on the sacrum in its fixated condition.

**Table 3.4** Bony specimens and/or joint data that have been excluded from the analysis

## CHAPTER 4: RESULTS

### 4.1 Stability of the different fixation techniques

Figure 4.1 shows the bending moment-angle characteristics of the three joints using the ensemble average as calculated over cycles 3 to 5 (i.e. Ens. Avg. and SD (cycles)), fixated with either the PMMA or the SOP technique. Table 4.1 shows the biomechanical parameters for the three joints of the bony spine specimens fixated with the two techniques, and the results of the non-parametric statistical tests. The table shows the results for the statistical comparison of the biomechanical parameters as measured between the two fixation techniques for all three joints in each of the specimens. The comparison of the biomechanical parameters between the three joints is also shown (this has already been mentioned).



**Figure 4.1:** Graph depicting the bending moment – angle characteristics of the PMMA and SOP fixated bony spines

**Table 4.1:** Biomechanical parameters of the three joints in the fixated spine specimens (Non-Parametric tests)

Biomechanical parameter	L5-L6		L6-L7	L7-S1
	PMMA (n=7)	SOP (n=9)	(n=7) (n=9)	(n=6) (n=9)
NZ(+) flex [deg]	PMMA : 0.23 ± 0.21 SOP : 0.36 ± 0.54 p-value : <b>0.5604</b>		0.48 ± 0.37 -0.063 ± 0.23 0.0070**	-0.088 ± 0.53 0.18 ± 0.35 <b>0.3458</b>
NZ(-) ext [deg]	PMMA : -0.14 ± 0.17 SOP : -0.019 ± 0.51 p-value : <b>0.7913</b>		0.092 ± 0.35 -0.18 ± 0.25 <b>0.1530</b>	-0.34 ± 0.43 0.0041 ± 0.35 <b>0.1255</b>
NZ total [deg]	PMMA : 0.37 ± 0.2 SOP : 0.38 ± 0.22 p-value : <b>0.9578</b>		0.39 ± 0.22 0.12 ± 0.069 0.0129*	0.26 ± 0.17 0.17 ± 0.16 <b>0.3458</b>
ROM(+) flex [deg]	PMMA : 5.9 ± 1.7 SOP : 5.2 ± 2.3 p-value : <b>0.7913</b>		6.7 ± 2.4 0.44 ± 0.33 0.0009**	1.2 ± 0.82 0.69 ± 0.37 <b>0.2386</b>
ROM(-) ext [deg]	PMMA : -1.9 ± 0.57 SOP : -1.5 ± 0.76 p-value : <b>0.2664</b>		-0.54 ± 0.39 -0.4 ± 0.27 <b>0.4914</b>	-1.3 ± 0.84 -0.69 ± 0.42 <b>0.1255</b>
ROM total [deg]	PMMA : 7.8 ± 1.9 SOP : 6.7 ± 2.8 p-value : <b>0.7913</b>		7.3 ± 2.7 0.84 ± 0.31 0.0009**	2.5 ± 1.2 1.4 ± 0.51 0.0251*
EZS(+) flex [N.m/deg]	PMMA : 0.74 ± 0.49 SOP : 0.42 ± 0.85 p-value : <b>0.7913</b>		0.39 ± 0.47 0.21 ± 0.46 <b>0.6338</b>	0.52 ± 0.56 -0.34 ± 0.64 0.0184*
EZS(-) ext [N.m/deg]	PMMA : 0.36 ± 0.32 SOP : -0.091 ± 1.4 p-value : <b>0.9578</b>		-0.072 ± 0.77 0.033 ± 0.21 <b>0.9578</b>	0.2 ± 0.82 -0.21 ± 0.74 <b>0.4094</b>

p-values indicated in **Bold** are >0.05 and imply that the null hypotheses ( $H_0: \mu_{pmma} = \mu_{sop}$ ) cannot be rejected and is considered to be true.

\* p-value < 0.05: significant difference at 5% level; \*\* p-value < 0.01: significant difference at the 1% level



## 4.2 Strength of the fixation techniques

Failure of the fixation technique used in the present study was initially defined as any catastrophic failure of the implant, such as an abrupt loss of its ability to support the load. In order to detect failure, the measured bending moment versus time of the load ranges (3Nm, Ir0, Ir1, and Ir1p5) was inspected to identify any abrupt decreases in the bending moment. The tests using load ranges Ir0, Ir1 and Ir1p5 were all conducted in displacement control. Tests using load range 3Nm was conducted in force control. Note that only a complete loss of ability to support the load could be observed on the force controlled load range (3 Nm) as the force, and therefore the bending moment, was controlled.

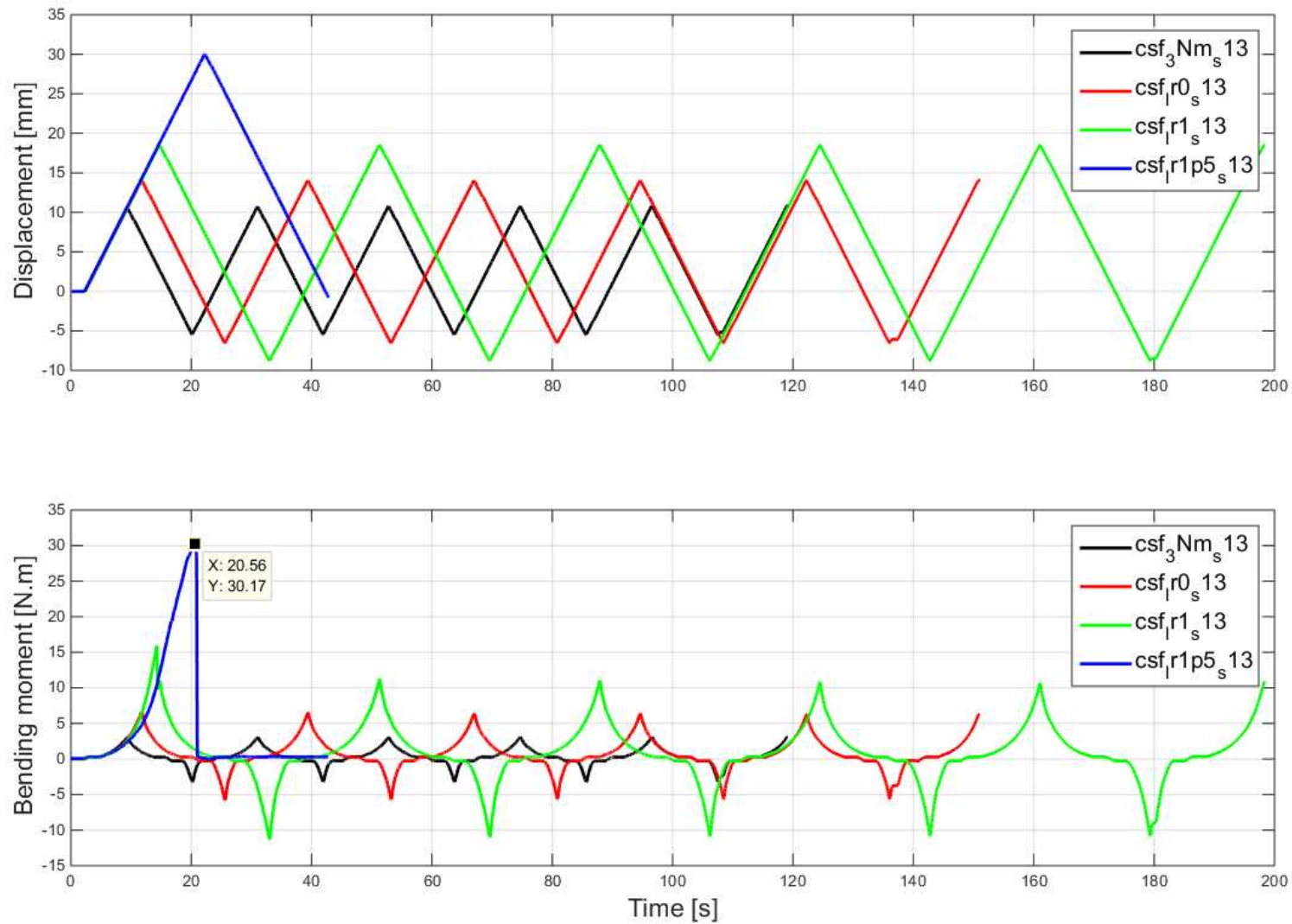
Figure 4.2 shows the displacement and bending moment versus the time for the load ranges for the PMMA fixated specimen No. 13. It is clear to see that an abrupt decrease in the bending moment occurred in load range 1.5 at 30.2 Nm. The mode of failure here was fracturing of the sacrum. Not all failures were associated with the same abrupt decrease in the bending moment as observed in Figure 4.2 for specimen No. 13.

In Figure 4.3 an abrupt decrease in the bending moment could not be observed, even with failure of the L6-L7 joint of specimen No. 1 with a load range of 1.5. What could be observed, however, was the fact that the peaks of the maximum deflection of the FBC-SLS and the maximum bending moment for Ir1p5, did not coincide. This is depicted in Figure 4.3 and could have been an indication of failure, as it was expected that the maximum deflection and maximum bending moment should coincide.

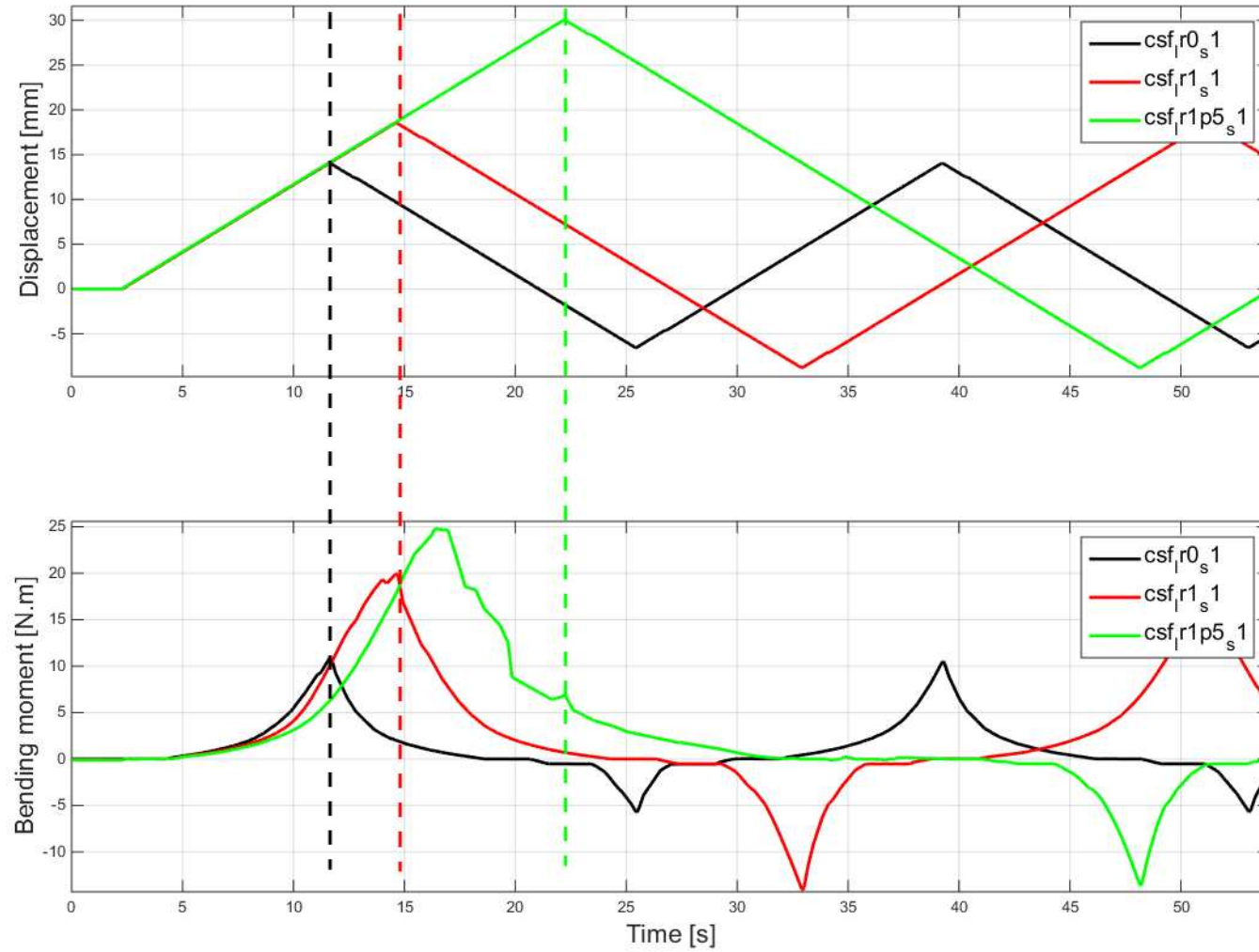
Therefore, in order to detect failure, the measured bending moment versus time was considered along with the displacement of the FBC-SLS of all the load ranges (3Nm, Ir0, Ir1, and Irp5) for each specimen. These plots were then inspected to identify any abrupt decreases in the bending moment and/or maximum bending moments which did not occur at maximum deflection of the FBC-SLS.



Table 4.2 and Table 4.3 present the load range in which failure occurred, the mode of failure, and the load at failure for the bony specimens that were fixated with the two specific fixation techniques. The failure modes are discussed in section 4.3.



**Figure 4.2:** Graphs depicting the displacement and bending moment vs. time of PMMA fixated for specimen 1.



**Figure 4.3:** Graphs depicting the displacement and bending moment versus time of PMMA fixated specimen 1.

**Table 4.2:** Load range at which the pin-PMMA fixated spine specimens failed

Specimen	Load range	Mode of failure (MoF) (see 4.3)				Bending moment at failure
		MoF1	MoF2	MoF3	MoF4	
s1*	lr1p5		X			24.83 N.m
s3	lr1p5				X	9.45 N.m
s6	lr1			X	X	6.97 N.m
s8	lr1p5			X		14.10 N.m
s9	lr1				X	5.37 Nm
s11*	lr1p5	X				20.25 N.m
s13	lr1p5			X		30.17 N.m
s16	lr1p5			X		FBC-SLS went outside of its max range in flexion and made contact with itself. Not able to get load at failure
s17	lr1p5			X		20.62 N.m
<b>Number of occurrences</b>		<b>1</b>	<b>1</b>	<b>5</b>	<b>3</b>	

\*Note that s1 and s11 had PMMA in the epoxy at the caudal end

**Table 4.3:** Load range at which the SOP fixated spine specimens failed

Specimen	Load range	Mode of failure (MoF) (see 4.3)				Bending moment at failure
		MoF1	MoF2	MoF3	MoF4	
s2	lr1p5			X		17.12 N.m
s4	lr1	X				21.13 N.m
s5	lr1			X		17.82 N.m
s7	lr0				X	10.27 N.m
s10	lr1	X				14.12 N.m
s12	lr1				X	11.7 N.m
s14	lr1				X	11.24 N.m
s15	lr1				X	17.39 N.m
s18	lr1p5	X				14.96 N.m
<b>Number of occurrences</b>		<b>3</b>	<b>0</b>	<b>2</b>	<b>4</b>	

### 4.3 Modes of failure (MoF)

Four modes of failure were detected during the testing of the fixated bony spines:

1. Failure of joint L5-L6. (Figure 4.4 & 4.5)
2. Failure of joint L6-L7 (Figure 4.7)
3. Failure (fracture) of the sacrum at the sacrum-epoxy interface. (Figure 4.9 & 4.10)
4. Failure of the connection between the epoxy and the sacrum. (Figure 4.12)

The failure modes are described below in terms of the physical failure and their associated characteristics, as observed on the displacement and bending moment measurements.

#### 4.3.1 MoF1: Failure of joint L5-L6

This mode of failure was associated with failure of the L5-L6 joint due to rupture of the associated joint capsules and the intervertebral disc, and an associated luxation of L5-L6.

Four specimens failed in this specific mode, of which three were SOP and one was a pin-PMMA specimen.



**Figure 4.4:** Photo of failure of the L5-L6 joint (pin-PMMA).

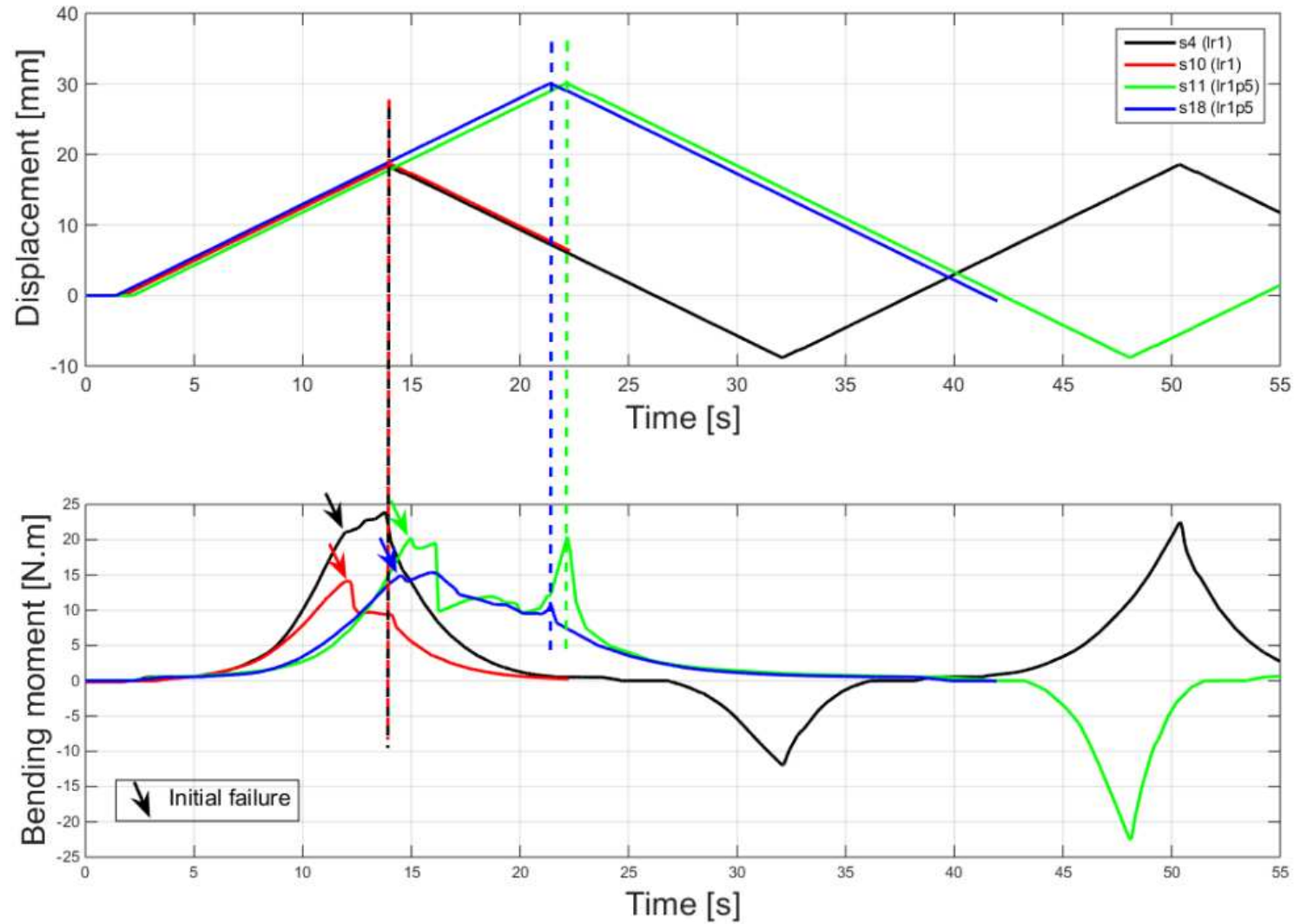


**Figure 4.5:** Photo of failure of joint L5-L6 in a SOP specimen.

Considering the bending moment verses time plot of specimen No.11 (see Figure 4.6), the initial failure (indicated by the green arrow) was not associated with an abrupt decrease in resistance to loading. The failed joint did, however, reach a point where there was an abrupt decrease in resistance to the applied load. A gradual decrease in resistance to load was observed at the point of failure for specimen No.18 (blue arrow). Specimen No.10 had an abrupt decrease in resistance to load that was associated with the point of failure (red arrow). Specimen No.4 also did not have an abrupt decrease in the resistance to load. Instead, a change in the incline of the bending moment was observed at the point of failure (black arrow).

After the point of failure of bony specimens No.11 and No.18, an increase in resistance was observed which corresponded with the maximum deflection in flexion of the spinal column. The reason for this second increase was not clear.





**Figure 4.6:** Graphs of failure of the L5-L6 joint (MoF1). Displacement and bending moment versus time of each specimen that had a failure of joint L5-L6.

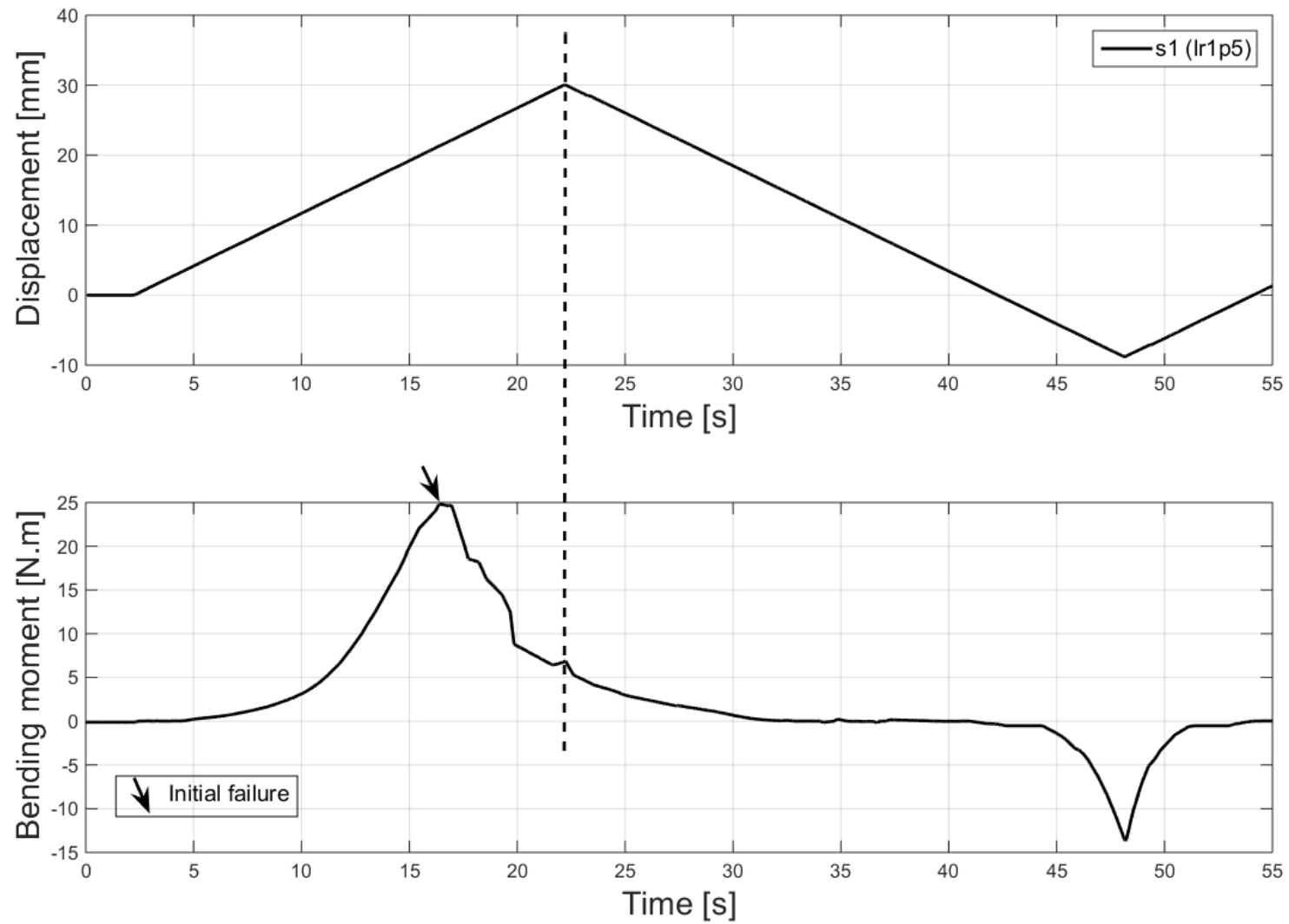
### 4.3.2 MoF2: Failure of joint L6-L7

This mode of failure was associated with failure of the L6-L7 joint due to rupture of the associated joint capsules and the intervertebral disc, as well as an associated luxation of the L6-L7 joint.



**Figure 4.7:** Photo of failure of joint L6-L7 joint in a pin-PMMA fixated bony specimen.

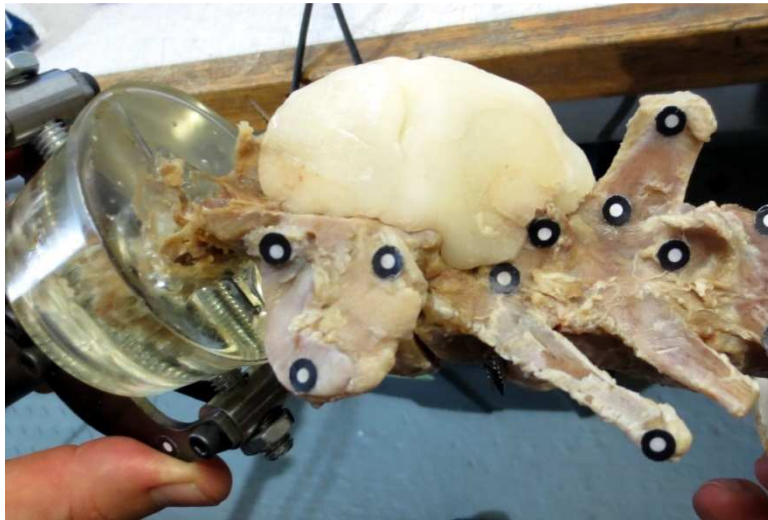
Considering the bending moment verses time plot of specimen No.1 (see Figure 4.8), the initial failure (indicated by the black arrow) was not associated with an abrupt decrease in resistance to loading.



**Figure 4.8:** Graphs of failure of the L6-L7 joint (MoF2). (Displacement and bending moment versus time)

### 4.3.3 MoF3: Failure (fracture) of the sacrum at the sacrum-epoxy interface

This failure was associated with a fracture of the sacrum at the sacrum-epoxy interface



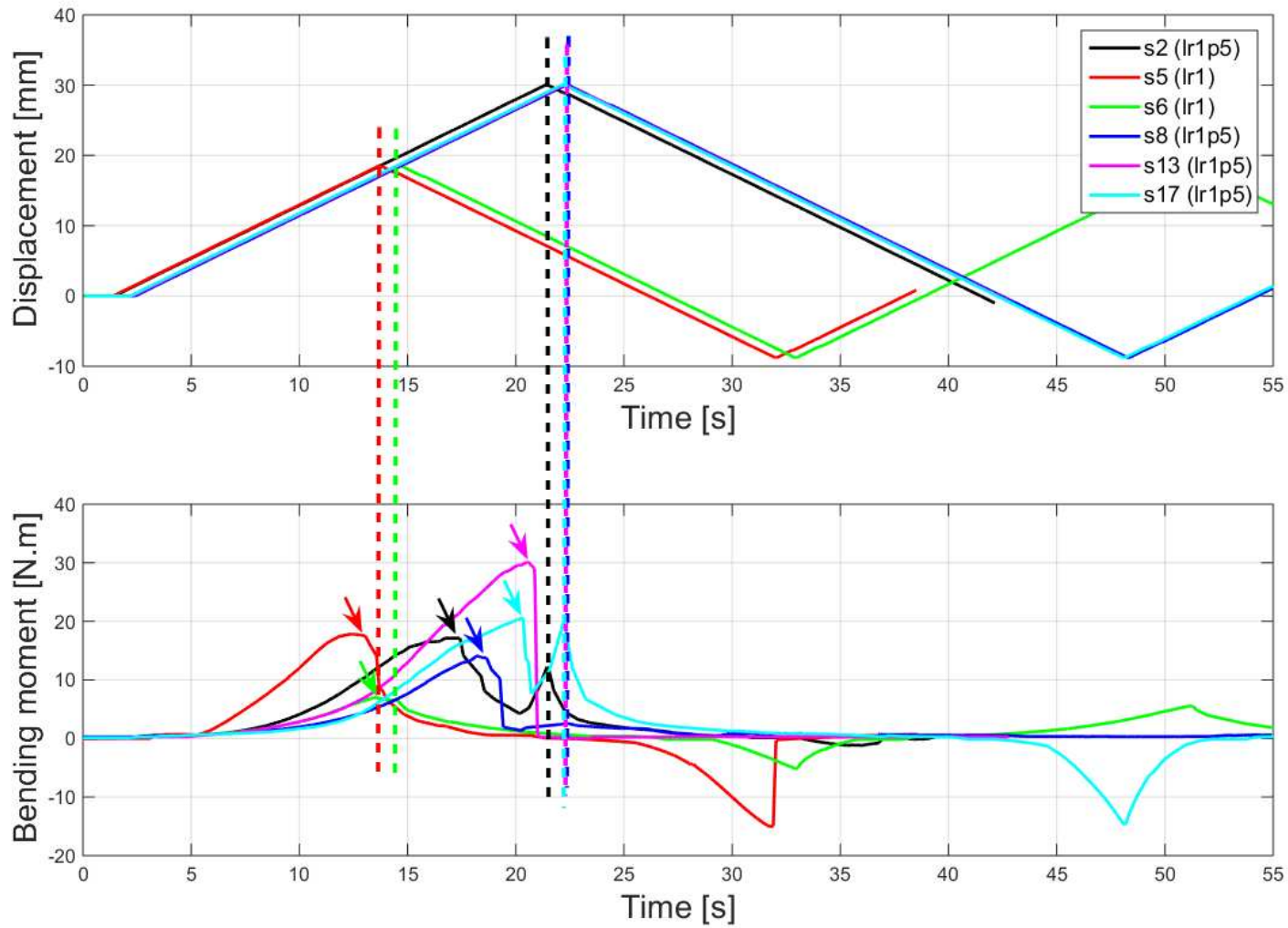
**Figure 4.9:** Photo of failure (fracture) of a sacrum at the sacrum-epoxy interface (pin-PMMA specimen).



**Figure 4.10:** Photo of failure (fracture) of a sacrum at the sacrum-epoxy interface (SOP specimen).

Considering the bending moment versus time plot of specimens No.2, 5, 8, 13, and 17 (see Figure 4.11), it can be noted that at the point of failure there was an abrupt decrease in resistance to loading. Specimen No.6 (indicated by the green line) also had an observable decrease in resistance at its point of failure, but it was not as abrupt when compared to the other specimens. Considering the results of specimen No.6 initially, it seemed to be a failure due to MoF3; but after careful analysis of the graphical data, it was reclassified as MoF4 and is discussed further under MoF4.

An increase in the resistance to loading was only observed after the point of failure of specimen No.17. This increase in resistance was due to contact that occurred between the fractured sacrum and the epoxy. Due to the fact that the FBC-SLS had gone outside of its maximum range in flexion while testing specimen No.16, it actually made contact with itself. This complication in the testing device unfortunately made it difficult to obtain the correct bending moment at failure.



**Figure 4.11:** Graphs of failure (fracture) at the sacrum-epoxy interface. (Displacement and bending moment versus time)  
 (Note that s6 was subsequently moved to MoF4)

#### 4.3.4 MoF4: Failure of the connection between the epoxy and the sacrum

This failure was associated with the breaking of the connection between the epoxy and the sacrum (epoxy-sacrum interface), due to the sacrum pulling out of the epoxy.



**Figure 4.12:** Photo of failure of the connection between the epoxy and the sacrum (MoF4).

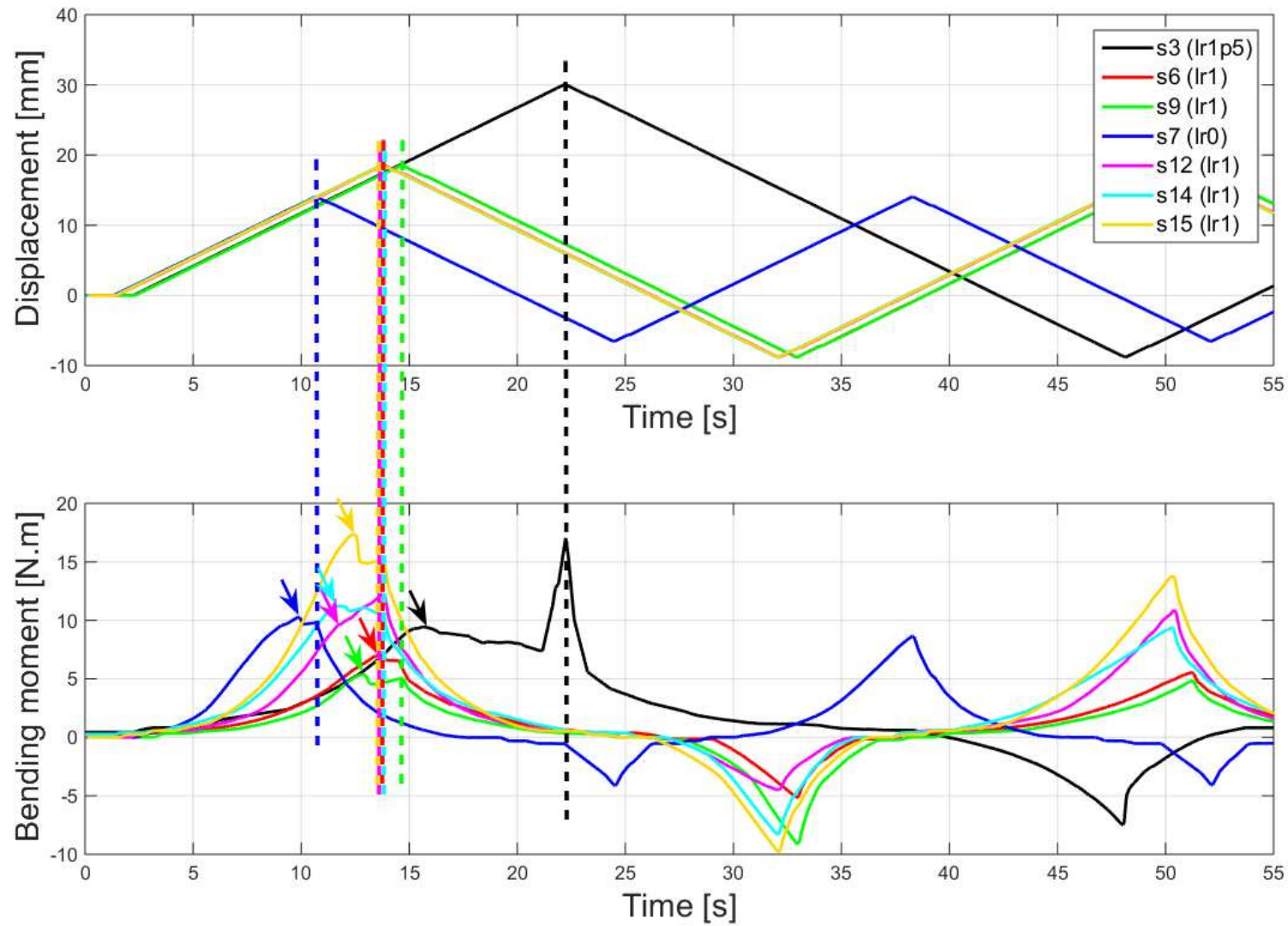
Considering the bending moment versus time plot of specimens No.3 and No.14 (see Figure 4.13), a gradual decrease in resistance to loading was observed at the point of failure. The reason for this second increase was not clear. Specimens No.6, 7, 9, and 15 had a more abrupt decrease in resistance to load associated with the point of failure than specimens No.3 and No.4.

The point of failure of specimen No.12 was not associated with any abrupt decrease in resistance to load. Instead, a change in the incline of the bending moment was observed at

the point of failure. The characteristic of this failure as observed for specimen No.12 was similar to that observed in specimen No.4.

After the point of failure of specimens No.3 and No.9, an increase in resistance was observed that corresponded with the maximum deflection of the bony specimen in flexion. This increase in resistance was much more pronounced for specimen No.3 than for specimen No.9.





**Figure 4.13:** Graphs of failure of the connection between the epoxy and sacrum. (Displacement and bending moment versus time).

# CHAPTER 5: DISCUSSION

## 5.1 Introduction

The aim of this study was to compare the mechanical behaviour of two internal spinal fixation techniques *in vitro* by measuring the bending moment-angle characteristic of the joint with the surgically simulated complete spinal fracture-luxation at L7, both in flexion and in extension within the elastic RoM of the bony spine segment. The hypothesis was that lumbosacral fracture-luxations stabilised with SOP plates would be as stable during flexion and extension as the conventional method with the use of pin-PMMA.

From the bending moment-angle characteristic, the spinal implant biomechanical parameters were extracted and used to quantify and compare the stability of the two different internal spinal fixation techniques. The strength of the two fixation techniques was quantified by the load sustained at failure in flexion.

Failure of each fixation technique was defined as any catastrophic failure of the implant; an abrupt loss of its ability to support the load. The rigidity and load sustained at the failure of the two fixation techniques were used to compare the strength of the two techniques.

## 5.2 Stability of fixation techniques

The results of the present study showed that there was no significant difference in the means of the biomechanical parameters of the injured lumbosacral joint between the two specific fixation techniques, except for the total RoM and the EZS in flexion. The evidence therefore supports the hypothesis that the stability of the injured lumbosacral joint between the two fixation techniques can be regarded as similar.

It should be noted that the results of the EZS were sensitive to the selection of the two points used to calculate the EZS, especially when the angles measured for the spinal joints were

small, such as in specimens with the fixated injured joints. Therefore, the statistical results of the EZS in flexion of the injured lumbosacral joint indicated that a difference between these two fixation techniques may be due to the manner in which the EZS was extracted, and not necessarily because of the physical differences between the two specific fixation devices.

The behaviour of the adjacent joints (L5-L6 and L6-L7) was compared between the two fixation techniques by using the same biomechanical parameters used for quantifying the stability of the injured lumbosacral joint. Joint L6-L7 was expected to show differences between the biomechanical parameters of the two fixation techniques, because this joint was included in the fixation with the SOP but not with the pin-PMMA technique.

As expected, the statistical results showed a significant difference in the mean values of the NZ and RoM in flexion and for the total of these parameters. However, with respect to extension, there was no significant difference between the NZ, RoM, and EZS. The intact joint L5-L6 had the same behaviour even though the spine was constrained differently by the two fixation methods. This could have been predicted, as joint L5-L6 should have the same characteristics, irrespective of the fixation technique, when subjected to the same bending moment. The deflection curve or stiffness of the entire bony spine specimen would, however, not be the same. The SOP resulted in the spine deviating more from its normal deflection curve or stiffness than with the pin-PMMA technique.

### **5.3 Strength of the fixation technique**

The strength of the two internal spinal fixation techniques was quantified by the load sustained at failure in flexion. The criterion set to define failure was an abrupt loss of its ability to support the load. However, many of the failures were not associated with an abrupt loss or ability to support the loading, as was shown in the section describing the modes of failure in the previous chapter. Therefore, in order to detect complete failure, the initial failure criterion was supplemented with the criterion that failure was assumed when the maximum



bending moment did not coincide with the maximum deflection of the load range and occurred before the maximum deflection.

By using these two specific failure criteria, the load at failure of the two different fixated bony specimens was obtained. There were four modes of failure. Most of the failures in both the pin-PMMA (seven out of nine) and SOP (six out of nine) fixated spines occurred in modes 3 and 4 (see Table 4.2

**Table**

**4.2**

and

**Table 4.3).** These two modes were associated with failure (fracture) of the sacrum at the sacrum-epoxy interface and/or at the connection between the epoxy and the sacrum.

Failure modes 3 and 4 could not be considered failures of the fixation technique itself, but rather a failure of the bony specimen due to boundary conditions imposed on it during the *in vitro* testing. It is therefore difficult to conclude the strength of the two fixation techniques or to make a comparison between the two. Inclusion of the pelvis in the fixation may also have resulted in a stronger epoxy-bone interface, but would have made monitoring of the L7-S1 joint more difficult due to superimposition of the ilium wing over the lumbosacral joint when viewed laterally. During the present study, the aim was to isolate the individual lumbar joints for evaluation and the decision was made to exclude the pelvis and sacro-iliac joint from the fixation. However, this resulted in a very small contact area between the sacral bone and epoxy on the caudal aspect of the specimen, leading to failure and the bone pulling out of the epoxy (MoF4), or excessive concentration of forces at the sacrum-epoxy interface and subsequent fracture of the sacrum (MoF3).

The different failure modes were analysed by considering the displacement and bending moment versus time. The characteristics of the failures were discussed in section 4.3 and, based on the observations, additional failure criterion was identified. Furthermore, a decrease in the peak bending moment was observed when comparing subsequent cycles in flexion and extension for many of the bony specimens. This change in the peak value was thought to be due to the connection between the epoxy and S1 not being completely rigid. However, after considering the bending moment of the pin-PMMA fixated spine specimen at S1, the decrease in peak bending moment values could not solely be attributed to this connection. Specimen No.1 had a significant amount of PMMA casted into the epoxy along with the first sacral vertebra (S1). This made the assumption of a rigid connection between the epoxy and S1 much more valid. Therefore, a decrease in the peak values may not only be due to the connection between the epoxy and S1, but may also be due to the viscoelastic behaviour of the spinal column and/or plastic deformation of, for example, the ligaments.



The importance of the biomechanical testing of any new fixation method and its comparison against the more conventional techniques currently in use cannot be overemphasised. If the surgeon knows both methods are equal in stability for fixating a fracture-luxation, other factors may then become more important in selecting a specific fixation method. Both techniques have advantages and disadvantages for spinal fixation that need to be considered.

The pin-PMMA method of stabilisation is flexible in terms of the number and position of the pins used. It can be applied at most levels of the vertebral column and is resistant to rotational forces<sup>24</sup>. Another advantage for the use of pin-PMMA is that this method can accommodate patients with a large variety of body weights, as the pin size and volume of PMMA needed to stabilise the pins can be adjusted<sup>51</sup>. This method also avoids compression of the spinal nerve roots<sup>24</sup>. This technique requires less soft tissue dissection for placement when compared to most other methods, and needs a shorter segment of vertebral column to be immobilised, thus allowing more normal spinal function<sup>6</sup>. The disadvantages of the pin-PMMA technique include thermal injury from the exothermic reaction during setting of the PMMA, pin migration, and increased risk of seroma and infection in the area<sup>23; 24</sup>. The bulk of the implant also makes closure of the wound more difficult<sup>49</sup>. The problem of pin migration or pin pull-out can be minimised with the use of partially threaded – preferably positive profile pins, instead of smooth pins<sup>6; 51</sup>. Due to the low cost and wide availability of pins or screws and PMMA, this method of stabilisation is commonly used in private practice

6; 8; 46; 51

For the SOP method of fixation, the manufacturer recommends the use of two bilaterally placed SOP plates applied to the dorso-lateral aspect of the vertebral bodies<sup>22</sup>. Longer plates are applied during fracture-luxations to engage two vertebrae on either side of the injury, with a minimum of three screws in the vertebral bodies on either side of the fracture-luxation<sup>22</sup>. This technique usually requires significant more soft tissue dissection for SOP placement, compared to pin-PMMA, and a longer segment of vertebral column is

immobilised. The use of dorso-lateral plating with SOP plates for fixation of the L7-S1 fracture-luxation is technically more demanding due to the relative plate versus vertebral body sizes. This may make it difficult to place the correct amount of screws cranial and caudal to the fracture line, and the ilium wings may prevent access to the vertebral bodies <sup>19</sup>.

An advantage of the SOP system is that the plate does not press down onto the underlying vertebral body bone, thereby accommodating irregularities of the vertebral column. The risk of damaging spinal nerves exiting the intervertebral foramina is reduced because the plate “stands off” from the vertebrae <sup>12; 21</sup>. Further advantages when compared to conventional plating methods include the use of standard cortical bone screws as locking screws, the ability to contour the plate in six degrees of freedom (two plane bending and torsion), high bending strength, and very competitive pricing due to the fact that costing is currently similar or less than conventional orthopaedic plating systems <sup>21; 22; 29</sup>. The cost of the SOP plate and specialised instrumentation needed are potential disadvantages when compared to the use of the considerably cheaper pin-PMMA technique. However, this is of little concern in the vast majority of cases where these types of luxations or fractures are fixed by experienced surgeons in specialized veterinary hospitals, where all the necessary equipment is available. Correct placement of the SOP plate is also technically more difficult when compared to the use of pin-PMMA for stabilisation of vertebral fracture-luxations.

## 5.4 Limitations of the study

The canine spine is anatomically best suited to withstand forces in flexion and extension since these forces are playing an important role in canine locomotion <sup>37</sup>. A study by Benninger *et al.*, (2004) showed that when flexion and extension are the main motions, the effects of lateral bending and rotation are actually very minimal <sup>7</sup>. Although the study of flexion-extension load testing is informative, it is not complete. Furthermore, only stability of the implants was considered in the present study. In order to have a comprehensive

biomechanical comparison, future work should also include strength evaluations as well as dynamic testing of these two fixation techniques under cyclic loading conditions.

Although the sample size of eighteen specimens was large enough for statistical analysis, a larger sample size is always more accurate and desirable.

The study was performed *in vitro* and the effect of bone healing played no role. The biomechanical testing was therefore almost exclusively directed at the implants.

The surrounding soft tissues that normally aid the stabilisation of fractures *in vivo* were removed. The validity of a spinal model without the external stabilising effects of the supporting musculature may be questionable, but it is likely that the load sharing ability of the lumbar muscles is drastically reduced in patients that are paralyzed due to spinal injuries anyhow<sup>43</sup>.

The three-dimensional movement of the lumbosacral spine is complex and studying movement in only one dimension (flexion-extension) is informative, but not complete.

The use of an oscillating saw to simulate the vertebral body fracture created a smooth cut surface and excluded any additional stability that could have been achieved by interdigitation of the bony fragments.

Although care was taken to standardise the implants, the absence of radiological abnormalities, body weight, and the loads applied to the specimens, variations such as potential interbreed differences and the structural properties of bone could not be standardised in this study.

Care was taken to minimise the “human factor” (i.e. the ability to precisely duplicate each osteotomy in terms of length and angle and placement of the implants), but it cannot be ruled out completely.





Although these factors could potentially influence the results, their impact is believed to be minimal.

## CHAPTER 6: CONCLUSION

The importance of biomechanical testing of any fixation method cannot be overemphasised. If the surgeon knows both methods are equal in stability and strength under static and dynamic loading conditions for fixation of a lumbosacral fracture-luxation, then other factors become more important in selecting a specific fixation technique.

Wilke *et al.*, (1998) stated that an estimate of the expected clinical success of an implant may be obtained by comparing the implant to those for which reasonable clinical experience is available<sup>50</sup>.

Statistically, this study did not show enough evidence to prove that there is a significant difference between the overall stability between the two methods of repair. Therefore, from the results obtained in comparing the two fixation techniques, the SOP method is expected to give the same clinical success – at least with respect to stability of the injured lumbosacral joint.

## CHAPTER 7: RECOMMENDATIONS

The following recommendations are made for future studies on the subject:

- Standardised material which is similar in structural and material properties to normal bone and uniform in size and shape should be used. If used as part of a physical spine model instead of canine cadaver bones, the results could potentially be more accurate.
- Different configurations of pin-PMMA can be used, incorporating the ilium – as described in some studies – and comparing different types of bone plates, such as DCP and LC-DCP.
- *In vivo* studies should be conducted to evaluate the clinical application of the results. *In vivo* studies have the added advantage of taking factors such as soft tissue support, the interdigitation of fractures, and bone healing into account; potentially rendering more meaningful results.
- The strength test considered in this study was merely static. Cyclic fatigue testing would involve the application of repeated (cyclic) loads on a specimen to simulate how it will perform during normal actual use<sup>39</sup>. Fatigue testing can be considered more physiologically accurate in terms of the forces that are acting on the spine construct during normal activity in the patient after stabilisation, and would therefore be able to produce a more complete picture of the strength of the specific fixation method.
- The spine model mentioned above would be useful in fatigue testing, as these specimens can be exposed to room temperature for longer periods without causing the degradation detected in a normal canine cadaveric spine specimen.
- Larger sample sizes can then be used to minimise statistical errors.

## REFERENCES

1. Agnello K A, Kapatkin A S, Garcia T C, Hayashi K, Welihozkiy A T, Stover S M (2010). Intervertebral biomechanics of locking compression plate monocortical fixation of the canine cervical spine. *Veterinary Surgery* 39: 991-1000.
2. Aguila A Z, Manos J M, Orlansky A S, Todhunter R J, Trotter E J et al. (2005). In vitro biomechanical comparison of limited contact dynamic compression plate and locking compression plate. *Veterinary and Comparative Orthopaedics and Traumatology* 18: 220-226.
3. Ahmad M, Nanda R, Bajwa A, Candal-Couto J, Green S, Hui A (2007). Biomechanical testing of the locking compression plate: when does the distance between bone and implant significantly reduce construct stability *Injury* 38: 358-364.
4. Amsellem P M, Egger E L, Wilson D L (2010). Bending characteristics of polymethylmethacrylate columns, connecting bars of carbon fiber, titanium, and stainless steel used in external skeletal fixation and an acrylic interface. *Veterinary Surgery* 39: 631-637.
5. Auger J, Dupuis J, Quesnel A, Beauregard G (2000) Surgical treatment of lumbosacral instability caused by discospondylitis in four dogs. *Veterinary Surgery* 29: 70-80.
6. Beaver D P, MacPherson G C, Muir P, Johnson K A (1996) Methyl-methacrylate and bone screw repair of seventh lumbar vertebral fracture-luxations in dogs. *Journal of Small Animal Practice* 37: 381-386.
7. Benninger M I, Seiler G S, Robinson L E, Ferguson S J, Bonél H M, Busato A R, et al. (2004) Three-dimensional motion pattern of the caudal lumbar and lumbosacral portions of the vertebral column of dogs. *American Journal of Veterinary Research* 65: 544-551.



8. Blass C E, Seim H B (1984). Spinal fixation in dogs using Steinmann pins and methylmethacrylate. *Veterinary Surgery* 13: 203-210.
9. Boero Baroncelli A, Reif U, Bignardi C, Peirone B (2013). Effect of screw insertion torque on push-out and cantilever bending properties of five different angle-stable systems. *Veterinary Surgery* 42: 308-315.
10. Bruce C W, Brisson B A, Gyselinck K (2008). Spinal fracture and luxation in dogs and cats: a retrospective evaluation of 95 cases. *Veterinary and Comparative Orthopaedics and Traumatology* 21: 280-284.
11. DeTora M, Kraus K (2008) Mechanical testing of 3.5 mm locking and non-locking bone plates. *Veterinary and Comparative Orthopaedics and Traumatology* 21: 318-322.
12. Downes C J, Gemmill T J, Gibbons S E, McKee W M (2009). Hemilaminectomy and vertebral stabilisation for the treatment of thoracolumbar disc protrusion in 28 dogs. *Journal of Small Animal Practice* 50: 525-535.
13. Dulisch M L, Nichols J B (1981). A surgical technique for management of lower lumbar fractures: case report. *Veterinary Surgery* 10: 90-93.
14. Early P, Mente P, Dillard S, Roe S (2013). In vitro biomechanical comparison of the flexion/extension mobility of the canine lumbosacral junction before and after dorsal laminectomy and partial discectomy. *The Veterinary Journal* 196: 533-535.
15. Fehlings M G, Perrin R G (2006). The timing of surgical intervention in the treatment of spinal cord injury: a systematic review of recent clinical evidence. *Spine* 31: S28-35; discussion S36.



16. Garcia J N P, Milthorpe B K, Russell D, Johnson K A (1994). Biomechanical study of canine spinal fracture fixation using pins or bone screws with polymethylmethacrylate. *Veterinary Surgery* 23: 322-329.
17. Haas S S, Brauer G M, Dickson G (1975). A characterization of polymethylmethacrylate bone cement. *Journal of Bone and Joint Surgery - American Volume* 57: 380-391.
18. Hente R, Lechner J, Fuechtmeier B, Schlegel U, Perren S (2001). Der Einfluss einer zeitlich limitierten kontrollierten Bewegung auf die Frakturheilung. *Hefte Unfallchirurg* 283: 23-24.
19. Jeffery N D (2010). Vertebral fracture and luxation in small animals. *Veterinary Clinics of North America, Small Animal Practice* 40: 809-828.
20. Johnson K (2009). Editorial: Locking plates - The ultimate implant ?. *Veterinary and Comparative Orthopaedics and Traumatology* 22: I-II.
21. Kowaleski M (2009). Locking plate systems: SOP and ALPS. Spring Scientific Meeting, Austin Court, Birmingham, 1st April 2009
22. Kraus K H, Ness M G, (2007). Orthomed Product information brochure: Standard operating procedures for SOP fixation of fractures. West Yorkshire: Orthomed. Available at [http://www.orthomed.co.uk/product-manuals/20130603100215\\_sop\\_sop\\_english\\_web.pdf](http://www.orthomed.co.uk/product-manuals/20130603100215_sop_sop_english_web.pdf) (accessed May 2015).
23. McAnulty J F, Lenehan T M, Maletz L M (1986) Modified segmental spinal instrumentation in repair of spinal fractures and luxations in dogs. *Veterinary Surgery* 15: 143-149.



24. McKee W M, (2009). SOP and Spines in Proceedings of the spring scientific meeting of the British Veterinary Orthopaedic Association, Austin Court, Birmingham, 1 April 2009.
25. Meheust P, Mallet C, Marouze C (2000). A new surgical technique for lumbosacral stabilization: arthrodesis using the pedicle screw fixation. Anatomical aspects. *Pratique Médicale & Chirurgicale de l'Animal de Compagnie* 35: 193-199.
26. Meij B P, Suwankong N, Veen A J, Hazewinkel H A W (2007). Biomechanical flexion-extension forces in normal canine lumbosacral cadaver specimens before and after dorsal laminectomy-discectomy and pedicle screw-rod fixation. *Veterinary Surgery* 36: 742-751.
27. Miller D L, Goswami T (2007) A review of locking compression plate biomechanics and their advantages as internal fixators in fracture healing. *Clinical Biomechanics* 22: 1049-1062.
28. Miller E I, Acquaviva A E, Eisenmann D J, Stone R T, Kraus K H (2011). Perpendicular pull-out force of locking versus non-locking plates in thin cortical bone using a canine mandibular ramus model. *Veterinary Surgery* 40: 870-874.
29. Ness M (2009) The effect of bending and twisting on the stiffness and strength of the 3.5 SOP implant. *Veterinary and Comparative Orthopaedics and Traumatology* 22: 132-136.
30. Ness M G (2009) Repair of Y-T humeral fractures in the dog using paired "String of Pearls" locking plates. *Veterinary and Comparative Orthopaedics and Traumatology*. 492-497.
31. Perren S M (1979) Physical and biological aspects of fracture healing with special reference to internal fixation. *Clinical orthopaedics and related research* 138: 175-196.

32. Perren S M (2003). Backgrounds of the technology of internal fixators. *Injury* 34: 1-3.
33. Perren S M (2001). Evolution and rationale of locked internal fixator technology: introductory remarks. *Injury* 32: 3-9.
34. Roe S C, Pijanowski G J, Johnson A L (1988). Biomechanical properties of canine cortical bone allografts: effects of preparation and storage. *American Journal of Veterinary Research* 49: 873-877.
35. Seim H B (2002). Fractures and Luxations of the Lumbosacral spine, in Fossum TW (ed) *Small animal surgery* (2nd Ed): pp 1312-1318.
36. Shores A, Nichols C, Koelling H A, Fox W R (1988). Combined Kirschner-Ehmer apparatus and dorsal spinal plate fixation of caudal lumbar fractures in dogs: biomechanical properties. *American Journal of Veterinary Research* 49: 1979-1982.
37. Smith G K, Walter M C (1988). Spinal decompressive procedures and dorsal compartment injuries: comparative biomechanical study in canine cadavers. *American Journal of Veterinary Research* 49: 266-273.
38. Smolders L A, Voorhout G, van de Ven R, Bergknut N, Grinwis G C M, Hazewinkel H A W, et al. (2012). Pedicle screw-rod fixation of the canine lumbosacral junction. *Veterinary Surgery* 41: 720-732.
39. Sod G A, Mitchell C F, Hubert J D, Martin G S, Gill M S (2008). In vitro biomechanical comparison of locking compression plate fixation and limited-contact dynamic compression plate fixation of osteotomized equine third metacarpal bones. *Veterinary Surgery* 37: 283-288.
40. Strauss E J, Schwarzkopf R, Kummer F, Egol K A (2008). The current status of locked plating: the good, the bad, and the ugly. *Journal of orthopaedic trauma* 22: 479-486.



41. Ullman S L, Boudrieau R J (1993) Internal skeletal fixation using a Kirschner apparatus for stabilization of fracture/luxations of the lumbosacral joint in six dogs. A modification of the transilial pin technique. *Veterinary Surgery* 22: 11- 17.
42. Wagner M, Frenk A, Frigg R (2007). Locked plating: biomechanics and biology and locked plating: clinical indications. *Techniques in Orthopaedics. Locked Plating: Biomechanics and Biology and Clinical Indications, Part II* 22: 209-218.
43. Walker T M, Pierce W A, Welch R D (2002). External fixation of the lumbar spine in a canine model. *Veterinary Surgery* 31: 181-188.
44. Walter M C, Smith G K, Newton C D (1986). Canine lumbar spinal internal fixation techniques a comparative biomechanical study. *Veterinary Surgery* 15: 191-198.
45. Watine S, Cabassu J P, Catheland S, Brochier L, Ivanoff S (2006). Computed tomography study of implantation corridors in canine vertebrae. *Journal of Small Animal Practice* 47: 651-657.
46. Weh J M, Kraus K H (2007). Use of a four pin and methylmethacrylate fixation in L7 and the iliac body to stabilize lumbosacral fracture-luxations: a clinical and anatomic study. *Veterinary Surgery* 36: 775-782.
47. Wheeler J L, Cross A R, Rapoff A J (2002). A comparison of the accuracy and safety of vertebral body pin placement using a fluoroscopically guided versus an open surgical approach: an in vitro study. *Veterinary Surgery* 31: 468-474.
48. Wheeler S J (1986). Surgical conditions of the canine spine. *British Veterinary Journal* 142: 95-115.
49. Wheeler S J, Sharp N J H, (2005). Trauma in Sharp NJH, Wheeler SJ (eds): Small animal spinal disorders, diagnosis and surgery (ed 2). Edinburgh, Elsevier Mosby, pp 281-318.



50. Wilke H J, Wenger K, Claes L (1998). Testing criteria for spinal implants: recommendations for the standardization of in vitro stability testing of spinal implants. *European Spine Journal* 7: 148-154.
51. Wong W, Emms S (1992). Use of pins and methyl-methacrylate in stabilisation of spinal fractures and luxations. *Journal of Small Animal Practice* 33: 415-422.



## Appendix 1: Protocol Approval by the Research Committee.

Ref: V010/13

9 May 2013



University of Pretoria

**Faculty of Veterinary Science**  
Private Bag X04  
Onderstepoort  
0110

Tel: +27 12 529 8000

Fax: +27 12 529 8300

Prof GL Coetzee  
Department Companion Animal Clinical Studies  
([louis.coetzee@up.ac.za](mailto:louis.coetzee@up.ac.za))

Dear Prof Coetzee

### **PROTOCOL V010/13: BIOMECHANICAL STUDY TO COMPARE THE USE OF FOUR PIN AND POLYMETHYLMETHACRYLATE (PMMA) TO THE “STRING OF PEARLS” (SOP) INTERLOCKING PLATE SYSTEM TO STABILIZE CANINE LUMBOSACRAL FRACTURE-LUXATIONS – Dr JJ Nel**

I am pleased to inform you that the abovementioned protocol was approved by the Research Committee.

Kindly note that, if there are animal ethical issues involved in the project, the protocol needs to be approved by the Animal Ethics Committee as well before you may commence with the project.

Please take note of the attached document.

Kind regards

**NIESJE TROMP**  
SECRETARY: RESEARCH COMMITTEE

Copy: Prof JAW Coetzer, Deputy Dean: Research ([koos.coetzer@up.ac.za](mailto:koos.coetzer@up.ac.za))  
Dr JJ Nel, Researcher ([theflyingvet@gmail.com](mailto:theflyingvet@gmail.com))  
Prof JP Schoeman, HOD ([johan.schoeman@up.ac.za](mailto:johan.schoeman@up.ac.za))  
Ms M Human, Student Administration ([magda.human@up.ac.za](mailto:magda.human@up.ac.za))  
Ms Elmarie Mostert, Animal Ethics Committee ([elmarie.mostert@up.ac.za](mailto:elmarie.mostert@up.ac.za))



**Appendix 2: Protocol Approval by the Animal Ethics Committee.**



UNIVERSITEIT VAN PRETORIA  
UNIVERSITY OF PRETORIA  
YUNIBESITHI YA PRETORIA

## Animal Ethics Committee

PROJECT TITLE	Biomechanical Study to Compare the use of Four Pin and Polymethylmetacrylate (PMMA) to the "String-Of-Pearls" (SOP) Interlocking Plate System to Stabilize Canine Lumbosacral Fracture-Luxations	
PROJECT NUMBER	V010-13	
RESEARCHER/PRINCIPAL INVESTIGATOR	Dr. JJ Nel	

STUDENT NUMBER (where applicable)	90222289	
DISSERTATION/THESIS SUBMITTED FOR	MMedVet	

ANIMAL SPECIES	Canine	
NUMBER OF ANIMALS	20	
Approval period to use animals for research/testing purposes	June 2013–June 2014	
SUPERVISOR	Prof GL Coetzee	

**KINDLY NOTE:**

Should there be a change in the species or number of animal/s required, or the experimental procedure/s - please submit an amendment form to the UP Animal Ethics Committee for approval before commencing with the experiment

<b>APPROVED</b>	Date <b>21 June 2013</b>
CHAIRMAN: UP Animal Ethics Committee	Signature 



## Appendix 3: Approval from National Council of SPCA

**NATIONAL  
COUNCIL OF  
SPCAs**

Incorporated Association not for gain



**NASIONALE  
RAAD VAN  
DBVs**

Ingenylke-Vereeniging sonder winsdoelmerk

Reg. No. 003-189 NPO

Dr. JJ Nel  
PO Box 30821  
Wonderboompoort  
Pretoria

30 May 2013

Dear Dr Nel

**Re: Biomechanical Study To Compare The Use Of Four Pin And Polymethylmethacrylate (Pmma) To The "String-Of-Pearls" (Sop) Interlocking Plate System To Stabilize Canine Lumbosacral Fracture-Luxations.**

The National Council of SPCAs (NSPCA) hereby confirms that the Board of the NSPCA has approved your request for twenty (20) dog cadavers weighing between 30 – 35kg as per Rule 6.12 of the SPCA Rules, which reads, "A Society shall not supply live or dead animals to any organisation, body or person for research or teaching purposes. Trials may be conducted on live animals where such trials are for the benefit of the same species, cause no suffering, are done upon the premises of the Society, and are conducted under the supervision of as veterinarian approved by the Council. Further, prior to the conduct of such trial, the prior written consent of the Board must first be obtained, and the trial shall be conducted upon such terms and conditions as the Council deems fit."

The following conditions are set:

1. The cadavers that are to be used must be stray animals.
2. The carcass of the animal must be incinerated by Envirocin as per your email dated 24 May 2013 to Este Kotze – Deputy CEO of the NSPCA.
3. Animal Ethics Committee approval letter/certificate needs to be sent to the office of the NSPCA for record purposes.
4. It is the prerogative of the SPCA that you approach to decide if they are willing to assist with this request or not.
5. A letter needs to be sent to the office of the NSPCA stating the total number of animals obtained as well as the SPCA from where these animals were obtained.

Yours sincerely

**ESTE KOTZE**  
Deputy CEO  
National Council of SPCAs



## Appendix 4 Copyright permission



UNIVERSITEIT VAN PRETORIA  
UNIVERSITY OF PRETORIA  
YUNIBESITHI YA PRETORIA

Department of Library Services

Lecturer's Name : ANTOINETTE LOURENS  
Department : COMPANION ANIMAL SURGERY  
Course code : CHR 882  
No. of students : 0  
Dissemination : Electronic Copy

Copyright permission has been granted (2016) for the reproduction of extract(s) from the publication(s) as per schedule for Electronic Copy.

Book/Journal Title	Author/Editor	Chapter/Article title	Author	ISBN/ISSN	Publisher	Page numbers
EUROPEAN SPINE JOURNAL	H.-J. WILKE, K. WENGER, L. CLAES	TESTING CRITERIA FOR SPINAL IMPLANTS: RECOMMENDATIONS FOR THE STANDARDIZATION OF IN VITRO STABILITY TESTING OF SPINAL IMPLANTS. FIG 3	H.-J. WILKE, K. WENGER, L. CLAES	0940-8719 (PRINT) 1432-0832 (O)	SPRINGER	P. 160

Please note that this permission is subject to the following condition:

1. The copied extract(s) may only be distributed by University of Pretoria and only to students enrolled for the specific course of study
2. Should the University of Pretoria decide to sell the copied extract(s) to the students enrolled for course of study, the selling price per unit shall not exceed the cost per unit incurred by University of Pretoria

Department of Library Services  
Maramba Library  
University of Pretoria  
PRETORIA, 0002  
Republic of South Africa

+27 (0)12 420 5575/6  
+27 (0)12 362 5100

[www.library.up.ac.za](http://www.library.up.ac.za)