Developing a methodology to perform measurements of the multi-spinal regions and lumbar-hip complex kinematics during dominant daily tasks

Raee Saeed Alqhtani

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School of Engineering, Cardiff University
Cardiff University, Cardiff, Wales, UK.

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Declaration

This work has not been submitted in substance for any other degree or --award at this or any other university of learning and is not being submitted concurrently in candidature for any degree or other awarded.

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Abstract

Introduction: Quantitative data of spinal range of motion in vivo is essential to improve clinicians' understanding of spinal pathologies, procedure of assessment and treatment. Accurate knowledge of physiological movement of lumbar spine regions, hip and the behaviour of each regional movement is important. Spine and hip motion play an essential role in daily functional activities, such as self-caring or performing occupational duties. Measuring the regional breakdown of spinal motion in three planes and describing the relative motion of different regions of the thoracolumbar (TL) spine can provide useful clinical information, which can be used in clinical procedure for spinal assessment. The relationship between the forward flexion (i.e. cardinal motion) and more functional tasks, such as lifting, stand-to-sit and sit-to-stand, as well as dividing the lumbar spine into more than one region, relative to the hip during these tasks, have not yet been established. Measuring the regional breakdown of spinal motions in three planes, as well as the relationship between lumbar spine and hip in sagittal plane, requires a multi-regional analysis system.

Aims and objectives: The fundamental aim was to explore range of motion and velocity magnitudes in flexion, extension lifting, stand-to-sit and sit-to-stand tasks, using three lumbar regions relative to the hip, and to determine correlations and differences between flexion and other dominant functional tasks. An objective was to obtain an appropriate measurement system that is capable of measuring dynamic movement in 'real time' and examine its validity against a "gold standard" system and its reliability, by measuring the range of motion of multi-spinal regions. Also, to demonstrate the relative contribution of five regions from within the thoracolumbar and head-cervical regions in 3D.

Methods: The selected system (tri-axial accelerometer sensors-(3A sensors)) was validated against a "gold standard" system (roll table (RT)) to demonstrate a correlation and root mean square errors (RMSEs) between the two devices. Reliability of the 3A sensors and the contribution of multi-spinal regions was assessed on 18 healthy participants. Two protocols were applied: in protocol one, two sensors were placed on the forehead and T1, to measure cervical ROM; in protocol two, six sensors were placed on the spinous processes of T1, T4, T8, T12, L3 and S1 to measure thoraco-lumbar regional range of motion. It also divided the lumbar spine as one single joint (S1 to T12) and as two regions (the upper (T12-L3) and lower (L3-S1)) and hip region. Data was gathered from 53 participants using four sensors attached to the skin over the S1, L3, T12 and lateral thigh. Two different statistical analyses were applied: one for analysing each particular region's contribution, relative to the hip; and another to analyse the correlation between the kinematic profiles of flexion and three sagittally dominant functional tasks (lifting, stand-to-sit and sit-to-stand).

Results: Validation of 3A accelerometer sensors system against the roll table revealed a strong correlation between the two systems average (ICC=.998 (95% CI=.993-.999)) and an acceptable rate of errors ranged from (2.54º (0.70%) to 5.01º (1.39%). It also demonstrated the reliability of this system, when the ICC values for all regions were high with relatively small errors associated with a novel multi-regional clinical spinal motion system. The ICC values for all regions were found to be high, ranging from .88 and .99 with 95% CI ranged from .62 to .99 while errors values ranged from 0.4 to 5.2°. The additional movement information, gathered from a multi-regional breakdown, adds insight into the relative contributions to spinal movement. Significant differences existed between ROM of LLS and ULS across all movements ($p<0.05$). A significant difference also existed between ULS-hip and LLS-hip ratio for the majority of tasks ($p<0.05$), and between ULS and LLS velocity for the majority of tasks ($p<0.05$). The findings from the lumbar spine as one region, underestimates the contribution of the lower lumbar and overestimates the contribution of the upper lumbar spine. Strong correlations for ROM are reported between forward flexion tasks and lifting for the LL spine ($r = 0.83$) and all regions during stand-to-sit and sit-to-stand ($r = 0.70$-0.73). No tasks were strongly correlated for velocity ($r = 0.03$-0.55).

Conclusion: The validity and reliability of the accelerometer sensors system is evidence of its ability to measure spinal movement. Since it is inexpensive, small, portable and relatively easy to use, it could be a preferable system for clinical application. The data, from multi-spinal regions, provides a novel method for practitioners to focus on a greater number of regions, rather than measuring only the three main areas of the spine (cervical, thoracic and lumbar). Investigating the lumbar spine as only one region risks missing out important kinematic detail. Further, the methodology provides the potential to measure functionally unique kinematics from more complex functional tasks, rather than generalised findings from clinical assessments of simple flexion.

Acknowledgments

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Finally, I dedicate this thesis to my parents who are always praying for me and wish me every success. My sincere prayers go to my father, who devoted his life to provide me with a happy life and an excellent education. My great thanks are due to my beloved mother, who knows what she means to me, for her emotional support and praying. I ask Allah to give them good health and happiness. I would like to thank my wife and sons for their unlimited patience and sacrifices. Without their unconditional love and support at every stage of my life, the completion of this thesis would not have been possible.

List of publications

Published Papers:

1. **Alqhtani R**, Jones M, Theobald P and Williams J. 2015. Reliability of an Accelerometer-Based System for Quantifying Multiregional Spinal Range of Motion. Journal of manipulative and physiological therapeutics, 38(4),pp 275–81.
2. **Alqhtani R**, Jones M, Theobald P and Williams J. 2015. Correlation of lumbar-hip kinematics between flexion and other functional tasks. Journal of manipulative and physiological therapeutics, 38 (6),pp 442–447.

3. **Alqhtani R**, Jones M, Theobald P and Williams J. 2015. Investigating the relative movement of upper and lower lumbar spine in everyday tasks. Journal of Manual Therapy, 0 (0),pp 0–17.

Conferences:

1. **Alqhtani R,** Jones M, Theobald P and Williams J. 2014. The reliability of novel multiregional spinal motion measurement device. International Journal of Therapy And Rehabilitation, 21, S6-S6.

2. **Alqhtani R,** Jones M, Theobald P and Williams J. 2014. Hip and lumbar motion: Is there a correlation between flexion and functional tasks? International Journal of Therapy and Rehabilitation, 21, S7-S7.

3. **Alqhtani R,** Jones M, Theobald P and Williams J. 2014. A novel method to evaluate the viability of 3A sensor measurements of primary motions for six cephalo-caudal regions and demonstrate range of motion for each particular region in 3D. In: International Conference on Spinal Manipulation, 25-27 October 2013, Phoenix, USA.

Prizes:

Prize for the Best Poster (**Alqhtani R**, Jones M, Theobald P and Williams J. 2014. Hip and lumbar motion: Is there a correlation between flexion and functional tasks? International Journal of Therapy And Rehabilitation, 21, S7-S7) awarded at 33rd Physiotherapy Research Society Spring 2014 Meeting on 14th May 2014 at the University of East Anglia

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Abbreviation

ROM	Range of motion
CROM	Cervical range of motion
RT	Rolly table
3A	Tri-Axial accelerometer sensors
AHRS	Attitude and heading reference systems
MRI	Magnetic resonance imaging
CMS	Coordinate measuring system
DSS	Department of Social Security
ALL	Anterior longitudinal ligament
PLL	Posterior Longitudinal ligament
MEMS	Micro-electromechanical systems
H-C	Head-Cervical
UT	Upper Thoracic
MT	Middle Thoracic
LT	Lower Thoracic
UL	Upper Lumbar
LL	Lower Lumbar
WL	Whole lumbar
T1	First thoracic vertebra
T4	Fourth thoracic vertebra
T8	Eighth thoracic vertebra
T12	Twelfth thoracic vertebra
L3	Third lumbar vertebra

S1	First sacral vertebra
3D	three-dimensions
IVD	Intervertebral Disc
MPU	Main Processor Unit
USB	Universal Serial Bus
TFL	Tensor Fascia Latae
PC	Personal Computer
MEMS	Micro-Electro-Mechanical Systems
OTS	Optical Tracking Systems
EMTS	Electromagnetic Tracking Systems
FG	Field Generator
IV	Impact Value
WV	Weighting Value
CN	Criteria Number
SD	Standard Deviation
ICC	Inter-class correlation coefficient
IMU	Inertial Measurement Unit system

Chapter 1: Thesis structure and Introduction

١.1 Thesis structure

Quantitative data of spinal range of motion in vivo is essential to improve clinicians' understanding of spinal pathologies, procedure of assessment and treatment. Accurate knowledge of physiological movement of lumbar spine regions, hip and the behaviour of each regional movement is important. Spine and hip motion play an essential role in daily functional activities, such as self-caring or performing occupational duties. Measuring the regional breakdown of spinal motion in three planes and describing the relative motion of different regions of the thoracolumbar (TL) spine can provide useful clinical information, which can be used in clinical procedure for spinal assessment. The relationship between the forward flexion (i.e. cardinal motion) and more functional tasks, such as lifting, stand-to-sit and sit-to-stand, as well as dividing the lumbar spine into more than one region, relative to the hip during these tasks, have not yet been established. Measuring the regional breakdown of spinal motions in three planes, as well as the relationship between lumbar spine and hip in sagittal plane, requires a multi-regional analysis system.

This thesis is structured into seven chapters and an appendix as follows:

Chapter 1: This chapter gives thesis structure and an introduction, outline on spinal disorders and how they affect people's lives in many ways including socially and economically. This chapter also highlights the value of the methodology used in this thesis and the possibility to apply this during clinical protocols. It outlines the motivation for the experimental studies of the thesis and contains its aims and objectives, hypotheses and contributions.

Chapter 2: This chapter contains the literature review, which is divided into the following subsections: the functional anatomy of lumbar spine and hip, contribution of multi-spinal regions, lumbar-hip biomechanics, lumbar-hip complex kinematics during dominant functional tasks in sagittal plane, and spinal motion measurement techniques. There is also a bullet-point summary of this chapter.

Chapter 3: This chapter contains the method applied in the selection process of the measurement system, programming methods, validity of study methods, and reliability of study methods, while it also divides the lumbar spine and correlation of the flexion movement along with other study methods.

Chapter 4: This chapter contains the results of the experimental studies. Various tables and figures of the four experimental studies have been presented in this chapter.

Chapter 5: This chapter contains a number of subsections, which discuss and compare the interoperability of the findings, procedures, limitations and applications.

Chapter 6: This chapter consists of recommendations for future work and conclusions of the experimental studies.

Chapter 7: This chapter contains a list of references.

Finally, there are four appendices. Appendix A contains the publications, which were produced from this thesis. Appendix B contains the participants' consent and information forms, Appendix C contains MatLab written programmes and figures and Appendix D contains tests for normal distribution and homogeneity of variance.

1.2 Introduction

Spinal mobility is an essential function for work duties and self-care during daily living activities. The average number of spinal movements in sagittal plane performed every 24 hours is approximately 4,400, with 66% of these movements ranging between 5° and 10° (Rohlmann et al. 2014). Spinal mobility is accomplished by a coordination of physiological and mechanical interaction between bones, joint articulations, ligaments and muscles which are controlled by central and peripheral nervous systems (Lebel et al. 2015). The spinal column's structure, functions and activities make it one of the most vulnerable body parts to injuries. The human spine is a mechanical system involving bone, facet joints, discs, ligaments and muscles. In such a system, the vertebrae may be considered as levers while the discs and facets act as pivots. Muscles and ligaments act as passive restraints and actuators, respectively (Esat 2006). Lower back pain influences the functions and kinematics of the spine and contributes to changing hip and lumbar spine mobility (Pearcy et al. 1985; Mellin 1990; Esola et al. 1996). Spinal disorders may often not be life-threatening, but they do represent a major public health problem, widespread among Western industrialised people (Deyo et al. 1998). The number of spinal disorders has increased significantly in recent years and more so than any other ordinary form of incapacity in Britain. From 1986 to 1992, back pain disability increased by 104%, while other causes of disability rose by 60% (Moffett et al., 1995). Furthermore, 116 million days of productivity were lost from 1994-1995 exclusively because of inability to work caused by back pain (DSS, 1998). Lumbar disorder is the most significant health and socioeconomic condition, which causes disability (Frymoyer 1988). The acute stage of lower back pain is one of the most common forms of the disorder (Ehrlich 2003) and is often connected with decreased lumbar and hip motion (Dolan and Adams 1993; Esola et al. 1996), followed by functional impairment (Cox et al. 2000). Lower back pain sufferers, on their way to chronic stages, are also limited in postural control, coordination and reaction time (Luoto et al. 1996).

Such disorders affect a vast number of communities annually, causing massive anxiety and economic hardship. The high prevalence of spinal problems is a strong reason for visiting orthopaedic and physiotherapy clinics, hospitals and other health care service centres. Furthermore, spinal disorders may cause disability and render sufferers unemployed and the considerable epidemiological and economic impact of spinal disorders on the public is expected to rise further (DSS, 1998). It's often the lower back pain lead to mobility impairment particular in sagittal plane associated with dominant daily functional takes on different forms such as difficulty in sitting down, rising, standing, walking and stair climbing. Therefore, an understanding of the lumbar spine kinematics is essential to recognise injury mechanisms and disorders in the spine in order to improve clinical service (Wang 2012). The complex structures, physiological loading and limitation in methods, means that an understanding of the in vivo biomechanics of the lumbar region is still limited (Wang 2012). Lower back pain is known to decrease the movement of the lumbar spine and hips (Esola et al., 1996) and such a problem could affect the number of functional forms (Cox et al., 2000). Lower back pain sufferers who complained about radiated pain in lower extremities display poor spine-hip coordination and considerable limitations of movement at lumbar spine and hip over sagittal tasks (Shum etal., 2005). This impairment may result from an increase in tissue stiffness, which can lead to a reduction in stretch tolerance of the hamstring muscles, while unusual tension in the sciatic nerve or its composing nerve roots may also be related (Goeken et al., 1991; Halbertsma et al., 2001).

Distinguishing the biomechanics of the hip is essential to improving the evaluation and treatment of several pathologic spinal conditions (Wong and Lee, 2004). Recognising the relationship between lumbar spine and hip kinematics is fundamental in order to facilitate the development of professional examination procedures and rehabilitation programmes (Lee and Wong, 2002). Similarly, identifying the relationship between hip and lumbar spine kinematics will improve the evaluation protocols of the joint function, the development of therapeutic programmes, procedures for planning reconstructive surgeries and the design and development of total hip prostheses (Johnston et al., 1998). The relative movement behaviour of the hip and its interaction with the lumbar spine has been suggested as being important (Lee and Wong, 2002; Sahrmann, 2002; O'Sullivan, 2005); therefore, understanding the relationship between hip and spine is significant in relation to this. Increasing such knowledge provides valuable information to physiotherapists in applying the appropriate scenario for assessing the lumbar-hip complex, treatment and follow-up protocols particularly when obtaining normative data for these regions. The measurement of human movement is encouraged by different goals in clinical practice, such as in comparing normal with pathological movements (Wong et al., 2007). Clinical studies have previously confirmed differences in the ratio between those with and without back pain (Shum et al., 2005; Shum et al., 2007) due to the affect bending and compressive stresses have on the lumbar spine (Dolan and Adams, 1993; Tafazzoli et al., 2014). Furthermore, investigating multi-regional lumbar spine versus hip movement across dominant daily functional tasks would significantly assist in achieving a better understanding of lumbar spine kinematics, especially when supplemented by multi-regional velocities (Shum et al., 2010).

The ratio of hip to lumbar movement is important to better understand the movement behaviour of the relative regions of the musculoskeletal system. Ratios associated with high amounts of lumbar motion relative to hip motion suggest a large contribution from the spine to the total motion. This may be associated with higher levels of activity which may in turn lead to problems associated with overuse and repeated end range movement. In contrast, those with relatively greater hip contribution will use the spinal segments less and thus have less demand for them.

A series of studies have previously focused on quantifying the relationship between the lumbar spine relative to hip, during everyday tasks (Shum et al.2007a; Wong and Lee 2004; Lee and Wong 2002; Shum et al. 2005a; Paquet et al.1994); however, in all cases the lumbar spine was only considered as a single region. More recently, authors have adopted multi-regional lumbar spine regions without regarding hip movement, across clinical populations (Williams et al., 2012; Williams et al., 2013) and healthy subjects (Leardini et al., 2011; Parkinson et al., 2013), identifying differences in regional contribution. The previous authors have suggested that the upper and lower lumbar spines display differences in their kinematic behavior and measuring lumbar spine as a traditional single 'joint' would fail to identify such subtleties and may, therefore, over simplify the description of movement. None of these studies confirmed how the relationship between hip and UL and LL spinal regions are functionally different yet. Identifying that the relationship between the hip and these specific lumbar regions is functionally different and unique may help to know the affected region of lumbar spine. For instance, if LL region affected, the ratio of LL-hip and UL-hip may produce no significant difference and that will be indicator to identify the local of injury because the normal ratio of LL-hip suggested to be greater than UL-hip.

It is not currently well understood to what degree the sagittal motions, such as forward flexion, are related to more daily functional tasks such as lifting, stand-to-sit and sit-to-stand. If there is no correlation, then using forward flexion as a basis for exploring sagittal movement behaviour would be flawed, potentially leading to erroneous clinical judgements and reasoning. It may be assumed, however, that forward flexion is closely related to other sagittal tasks, making the assessment of many tasks within the clinic unnecessary.

Previous research has suggested that the upper and lower lumbar (hereafter referred to individually as UL and LL, respectively) spine display differences in their kinematic behaviour and that measuring lumbar spine as a traditional single ‘joint’ would fail to identify certain subtleties and may, therefore, oversimplify the description of movement. To investigate the multi-regional movement in the lumbar spine-hip complex during dominant functional tasks and multi-spinal regions movement, a measurement system is required that is capable of measuring the movement in real time. Dynamic motion tracking over real time requires laboratory methods that are often complex, costly and constrained to the clinical setting. A portable motion analysis system, which overcomes these limitations and can be used in clinical examination protocols, is required.

1.2.1 Aims and Objectives

The aims of this thesis are the following:

Aim one: To review all existing technologies that capable for measuring spinal kinematics, and to identify that which offers the greatest potential for use in physiotherapy clinics. This system needs to be capable of tracking the movement of multi-spinal regions. Accordingly, three objectives were established to achieve this aim.

I. Establish a number of criteria to select the appropriate system that can capture the head-cervical, spine and hip kinematics.

II. Examine the validity of a selected system against a ’’gold standard’’ system.

III. Examine the reliability of the selected system by using a novel method which measure multi-spinal regions and head-cervical region.

Aim two: Quantify the relative contribution of five regions from within the thoracolumbar region as well as head-cervical region during flexion, extension, lateral flexion to right and to left, and rotation to right and to left in order to confirm the reliability of the selected system.

Aim three: Investigate whether dividing the lumbar spine as two separate regions with relative to hip will yield a different understanding of the movement behaviour compared with a traditional single joint region during the dominant functional tasks in sagittal plane. Hence, the following objectives will obtain by multi-regional analysis system to achieve this aim:

I. Exploring the range of motion magnitude using a traditional region of the lumbar spine as one single joint to compare this with a sectioned approach, where the lumbar spine is divided as two distinct regions, namely the upper lumbar and lower lumbar spine.

II. Exploring the velocity magnitude using a similar technique, used with range of motion, to demonstrate the movement behaviours. These regions will also be used to compare hip ratio, range of motion and velocity during these tasks.

Aim four: Investigate the correlation between the kinematic profile of lumbar-hip complex in flexion and three sagittally dominant functional tasks (lifting, stand-to-sit and sit-to-stand). In order to achieve this aim the following objectives were suggested:

I. Exploring the correlation between the range of motion of flexion and other dominant functional tasks for the anatomical regions of the upper and lower lumbar spine and hip. Exploring the correlation between the velocity of flexion and other dominant functional tasks for the anatomical regions of the upper and lower lumbar spine and hip.

1.2.2 Research hypothesises

This thesis contains four studies as following

1- Validity study which conducted to compare the similarities between new measurement system -i.e. tri-axial accelerometers sensors (3A)- and gold standard measurement system (Rolly Table)

2-Reliability study which conducted to determine the similarities between three trials and acceptable errors of measurements at different levels of spine during sagittal, frontal and transverse movements. This study also used to demonstrate contribution of multi-spinal regions and head-cervical region.

3- Study of measuring the multi-regional of lumbar spine comparing with hip.

4- Study of the correlation between the flexion movement and other sagittally-dominant functional movements will conducting.

1- <u>Validity study</u>

To examine the accuracy of the 3A sensors against a gold standard system (Rolly Table), concurrent validity used to compare the level of accuracy between two systems. Such types of validity are used to compare the measurement data obtained by a new measurement system, with measurement data which measured by a previously validated measure, often a gold standard measurement (Portney and Watkins, 2009).

Null Hypothesis

There is no correlation between orientation of 3A sensors and orientation of the gold standard system. When Pearson correlation (r) > 0.80 is reached, the null hypothesis will be rejected.

2- <u>Reliability study</u>

It is necessary to provide constant or reproducible values with tolerance errors of measurement when no variable is affecting the attribute that the measurement is quantifying (Rankin and Stokes 1998). The Intraclass Correlation Coefficient (ICC) will use to assess the reliability of a novel method which measures multi-spinal regions and head-cervical region in conducting an examination using a selected system.

Null Hypothesis

There is no correlation between three scores of multi-regional spine range of motion measured by a single rater (test re-test reliability) using a selected system. When the ICC of three measurements reaches > 0.80, the null hypothesis will be rejected.

3- **Study of measuring the multi-regional of lumbar spine comparing with hip.**

The hypothesis of this study are designed to investigate the relative movement of UL and LL spine in everyday tasks.

Therefore, this study sets out to investigate the following specific hypotheses:

Null Hypothesis 1

The LL region will not contribute to kinematics at flexion, extension, object lifting, stand-to-sit and sit-to-stand any more than at the UL region.

When the difference is significant (i.e. where the kinematics between LL and UL spine during these tasks is $(p \leq 0.05)$), the null hypothesis will be rejected.

Null Hypothesis 2

There is no difference in information between the kinematics ratio of LL-hip and ratio of UL-hip. When the difference is significant $(p \leq 0.05)$, the null hypothesis will be rejected.

4- **Study of the correlation between the flexion movement and other sagittally-dominant functional movements.**

The correlation between the kinematic profiles of flexion and three sagittally-dominant functional tasks (lifting, stand-to-sit and sit-to-stand) may not be sufficient to consider flexion as a basis for exploring sagittal movement behaviour and may lead to erroneous clinical judgements and reasoning.

Null Hypothesis

There is a correlation between flexion movement and other daily dominant tasks (i.e. lifting, stand-to-sit and sit-to-stand). When the difference is significant between flexion and any of these tasks *($p \leq 0.05$)*, the null hypothesis will be rejected.

1.2.3 Contributions

The motion analysis of human beings has been keenly researched in bioengineering and rehabilitation centres. The use of electronic sensors is regarded as being a potential method for human spinal motion analysis in clinical applications. The tri-axial accelerometer sensor system, which provides orientation and acceleration information with gravity orientation, was selected to use for measuring spine and hip kinematics in this research based on it being superior to other systems. Evidence from both validity and reliability studies of this system have confirmed its feasibility when conducting spinal measurement. It considers a viable option as it is small enough and sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics. Further, advantages which will add to physiotherapy practice are the following:

1- This thesis conducted new normative data describing the kinematics of multi-spinal regions. Therefore, physiotherapists are now capable to use these data as reference at similar protocol of assessment to identify abnormal movement. Known such data is crucial because analysing the range of motion of multi-spinal regions provides the opportunity to expand our perception regarding the severity of spinal disorders. For instance, the development of ankylosing spondylitis and the surgical influence on multiple-level discectomy or laminectomy (Hsu et al., 2008).

2- The benefit from findings of LL-hip and UL-hip ratio is that the physiotherapist capable now to use more detailed findings as a normative data during spinal regions assessment at physical therapy clinics. For instance, when the LL region has been affected, the ratio of LL-hip and UL-hip may produce no significant difference due to a decrease in both the LL ROM and velocity. Such a finding that the UL and LL are functionally independent is important for clinical practice to apply the treatment protocol on the affected region more than other and that save physiotherapist time and increase healing process and improve regional function.

3- The new information discovered that the sagittal kinematics of the hip and lumbar spine during trunk flexion are different from those observed during other dominant functional tasks in the same plane. This conclusion could change physiotherapist protocols of spinal assessment by adding more tasks such standing to sitting and sitting to standing and suggests that physiotherapists cannot rely on flexion assessment alone.

Chapter 2: Literature Review

2 Literature review

2.1 Functional anatomy of lumbar spine and hip

The purpose of this section is to provide an overview of the clinical anatomy of spinal regions in order to make the reader understands the nature of spinal curves and functions. The anatomy of the spine can be fully understood when its functions are considered first (Cramer & Darby 2013). The main functions of the spine are supporting the human in the upright position, allowing movement and locomotion and protecting the spinal cord, as well as various neurovascular structures (Middleditch & Oliver 2005). A basic understanding of the spine and hip anatomy and their functions are very important for physiotherapists to evaluate and treat spinal disorders.

The spine has four curves when viewed in the sagittal plane. The cervical and lumbar regions are convex anteriorly (lordotic), while the thoracic and sacral regions are convex posteriorly (kyphotic). The lordotic curves develop after birth as the infant's spine straightens out, which facilities development of the bipedal posture. Although there is a harmonious progression of these curves from one to the other, which may help to evenly distribute stress and strain, injuries occur more commonly at the junctional areas because of a difference in the relative stiffness of each anatomical segment of the spine (Middleditch & Oliver 2005).

2.1.1 Spinal vertebrae

The vertebral column normally consists of 24 separate bony vertebrae: cervical (7), thoracic (12) and lumbar (5) with these main regions described as mechanical structures (Figure 2.1.1). The lumbar spine region is connected by five fused vertebrae form the sacrum, which is connected with four fused vertebrae of the coccyx. The typical includes the vertebral body, pedicles, superior articular facet, inferior articular facet, transverse process, spinous process, vertebral foramen and lamina (Figure 2.1.2).

The lumbar spine is the anatomical region between the twelfth thoracic (T12) and the first sacral (S1) vertebrae (Burton et al. 1989) and it is considered a mechanical structure that works via a levers system, pivot activators and restraints (Wang 2012). The main functions of the lumbar spine are to allow range of motion in three-dimensions (3D), to provide lumbar stability and balance in either sagittal or coronal planes for upright position, and to bear the majority of weight of the trunk and upper limbs during body movement (Middleditch & Oliver 2005).

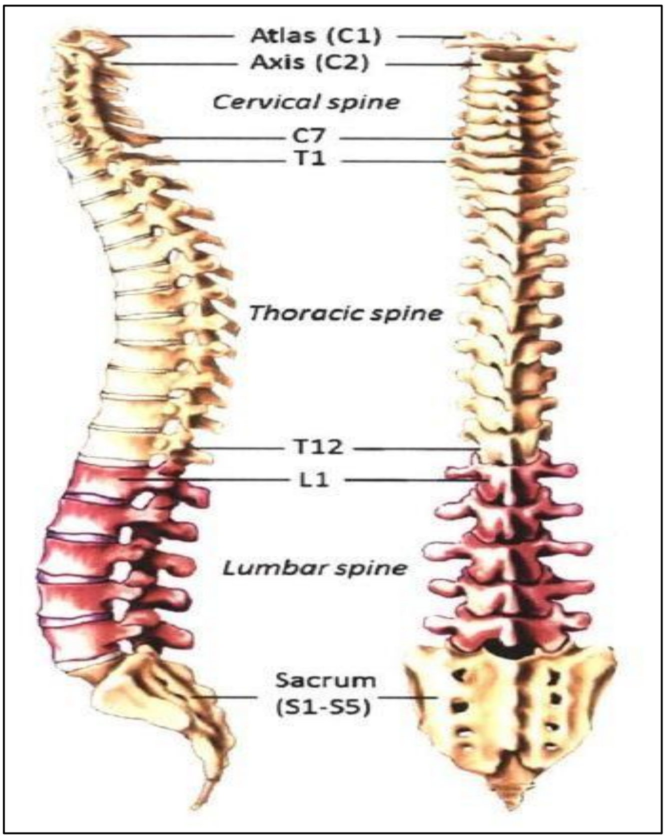


Figure 2.1.1: Lateral and posterior view of the spine (Wang 2012). Lateral view showing the cervical, thoracic, lumbar and sacral regions. Also, notice the cervical and lumbar lordoses and the thoracic and sacral kyphosis.

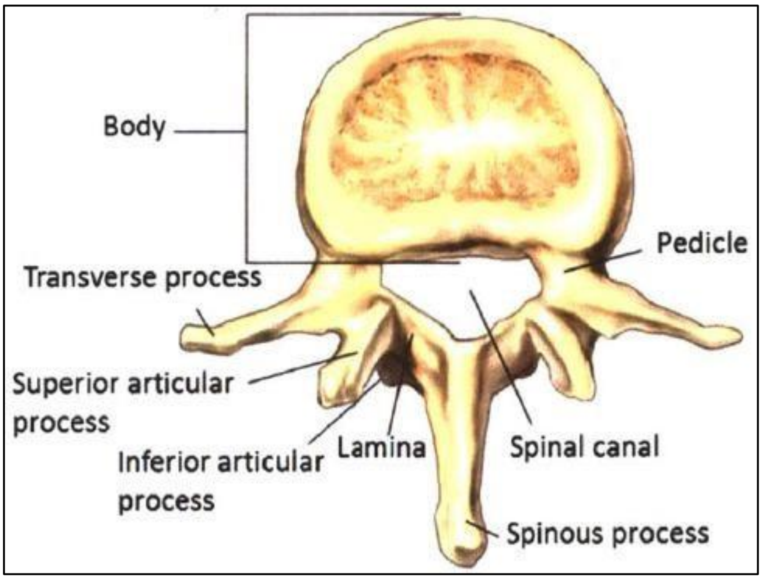


Figure 2.1.2: Superior view of lumbar vertebra (Wang 2012).

2.1.2 Spinal joints and intervertebral disc

The spinal vertebrae are formed as a column (i.e. each vertebra stacked on top of the other) (Figure 2.1.3). There are two spinal pillars, the anterior and posterior pillars. The anterior pillar consists of vertebral bodies and intervertebral discs. The principle functions of the anterior pillar are weight-bearing and shock-absorbing (Middleditch & Oliver 2005). The posterior pillar comprises the articular processes and epiphyseal joints (facets joints) which connect vertebrae together (Middleditch & Oliver 2005). Facet joints not only the structure which provides the articulation between vertebrae and firmness but there are other structures providing the stability for the vertebral column (Hazlett & Kinnard 1982). These structures work to provide the spine flexible and to transit great compressive loads. The spinal facet joints permit spine motion with four facet joints on each vertebra, two superior and two inferior.

The intervertebral discs are located between the vertebral bodies (with the exception of the first and second of the cervical vertebrae). They exhibit creep and relaxation behaviour and absorb pressure, distribute forces of weight and protect the vertebrae from grinding against each other (White and Panjabi 1990).

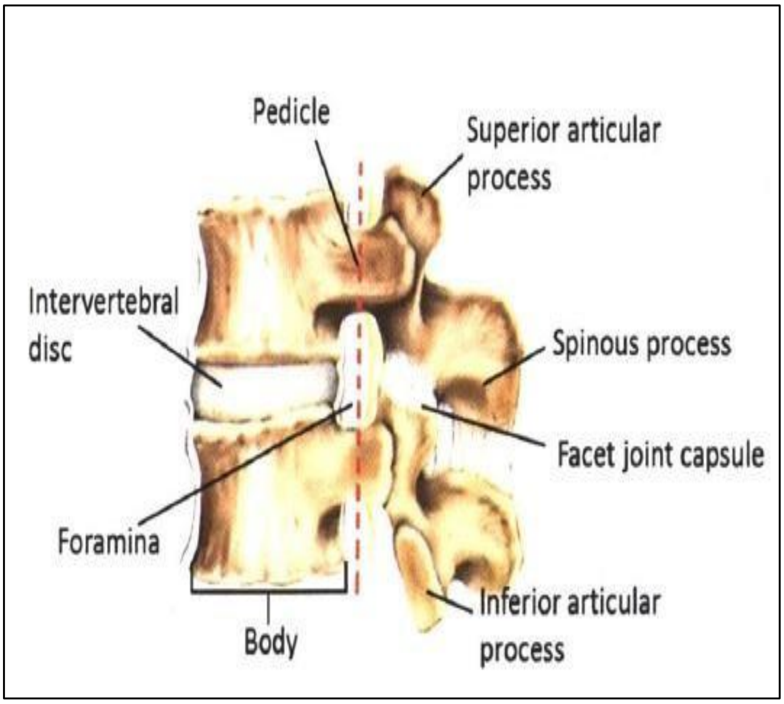


Figure 2.1.3: Section of two adjacent lumbar vertebrae, and the intervertebral disc separating the two vertebral bodies (Wang 2012).

2.1.3 Muscles, Ligaments and Tendons

Two major groups of muscles are working during flexion and extension. The spinal extensors work when the spine moves backwards, which allow us to stand up and lift objects, and are attached to the back of the spine. The spinal flexors work to move the spine forward and allow us to lift and control the arch in the lower back.

The vertebrae and discs are connected together by numerous ligaments. Ligaments articulate the bones to each other and work to permit tolerable physiologic motions, protecting the spinal cord and providing stability to the spine (White & Panjabi 1990). Muscles and muscle tendons influence the relative stability of joints as the function of the tendon is to connect muscle to bone. Indeed, muscles and tendons work to stabilise the vertebral spine by connecting the articulating bone ends together and preventing excessive movement in all directions (Hall 1999). The functions of strong ligaments are to link vertebrae together, to stabilise the spine and to protect discs. There are seven ligaments that connect one vertebra to the next: the anterior longitudinal ligament (ALL), posterior longitudinal ligament (PLL), the ligamentum flavum, facet capsular ligament, supraspinous ligament, interspinous ligament and intertransverse ligament (Figure 2.1.4). The ligaments contribute to the spine's stability is due to the position of each ligament that is dependent on its cross-section, its distance from the instantaneous axis of rotation and its orientation in space (White & Panjabi 1990).

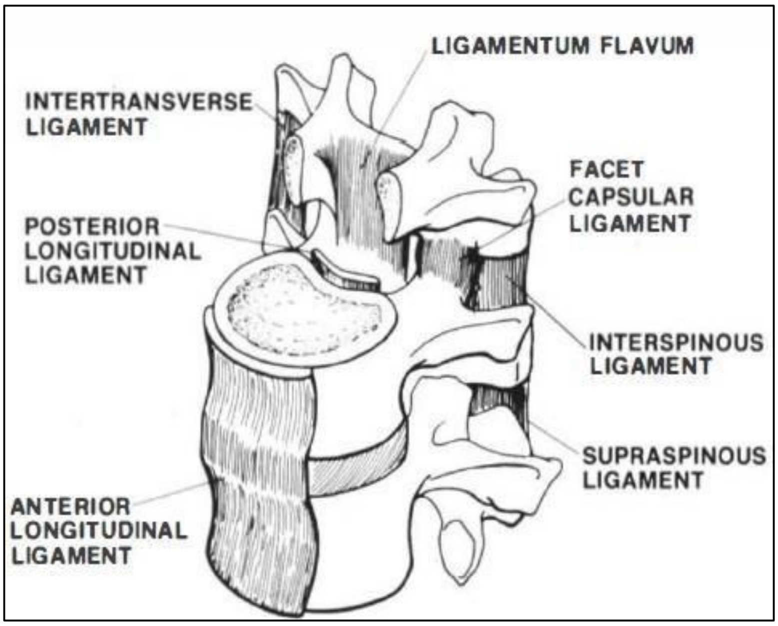


Figure 2.1.4: Spinal ligaments. Image adopted from White and Panjabi (1990).

2.1.4 Spinal cord and peripheral nerves

The spinal cord is surrounded by the relative spinal canal, composed of hard vertebrae accommodating the spinal cord and protecting it from injury (White and Panjabi 1990). Mechanically, the spinal canal decreases in length when the spine is extended and increases when the spine is flexed. Small nerve roots, which are called peripheral nerves, branch off from the spinal cord through spaces in between each vertebra and spread throughout the whole body (Figure 2.1.5). The spinal cord and the nerves are part of the central nervous system that includes the brain. In brief, the nerves are the neural message system of the body.

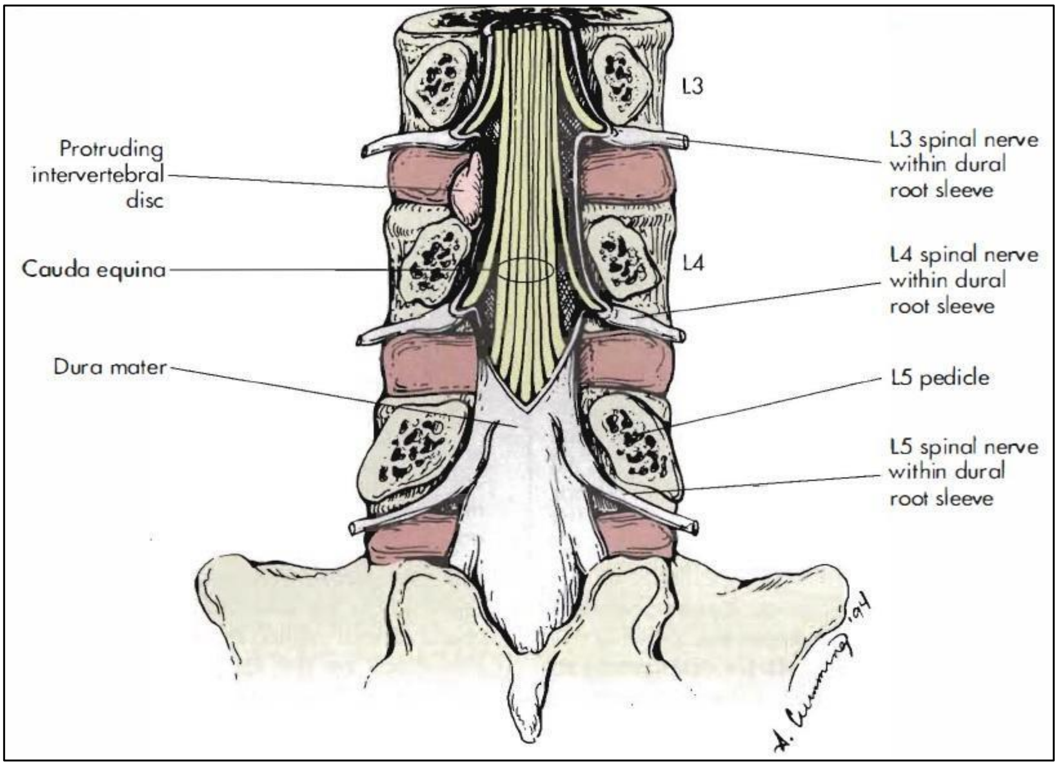


Figure 2.1.5: Spinal cord surrounded by the relative spinal canal and exiting nerve roots known as peripheral nerves (Cramer and Darby 2013)

2.1.5 Hip joints

In considering the hip kinematics, it is preferable to understand this joint as a stable ball-and-socket configuration in which the head of the femur and acetabulum can move in all directions such as the range of motion in sagittal uring flexion ranges from 0 to 140^{o} and extension from 0 to 15 (Middleditch & Oliver 2005). The hip joint is structured to allow for mobility in multiple directions and plays a fundamental role in stability (Figure 2.1.6). The hip joint allows for the movement of the lower extremity in three planes of motion: forward and, backward in sagittal plane, side to side in frontal plane and rotating right and left in transverse plane. The hip joint provides shock absorption to the thoracolumbar spine and upper body in addition to stability when in an upright position and during other weight-bearing activities. The hip joint is a classic ball-and-socket joint comprising four main components: bones, cartilage, ligaments and muscles.

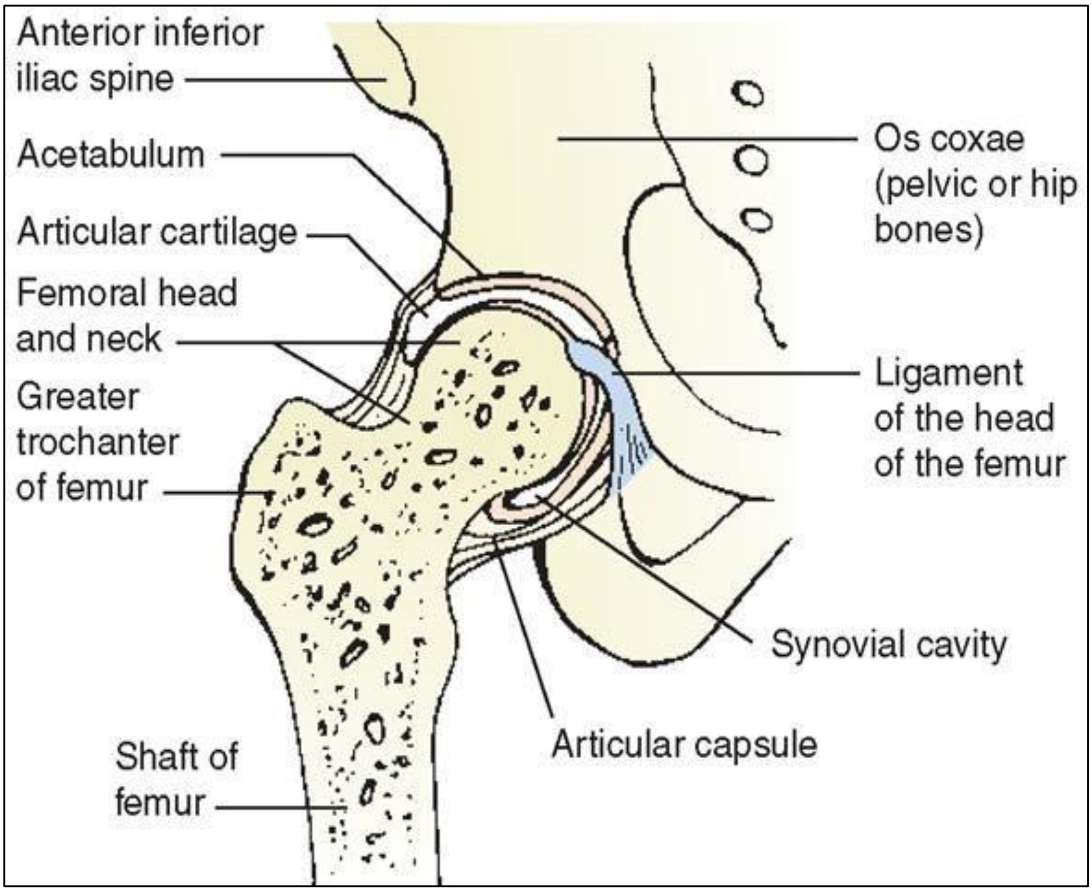


Figure 2.1.6: Hip joint. Section through right hip joint, showing insertion of head of femur into the acetabulum (http://what-when-how.com/nursing/the-musculoskeletal-systemstructure-and-function-nursing-part-3/).

2.1.6 Hip bones

Innominate bone forms the acetabulum in the shape of a cup (concave shape) with assistance from the ischium (about 40% of the acetabulum), ilium (40%) and the pubis (20%) (Schuenke et al 2006). Acetabulum labarum is a common location for tears which is an indication of hip arthroscopy (Byrd 2005). Even though such shape affects joint stability and constrains the hip joint mobility more than the glenoid labrum in the shoulder, it does serve its purpose. The hip joint connects two separate bones: the femur bone and the pelvis. The pelvis structure is two cup-shaded depressions called the acetabulum. The longest bone in the body (femur) adjoins the pelvis at the hip joint. The femur head is formed as a ball and fits closely into the acetabulum, forming the ball-and socket joint of the hip, allowing the lower extremity to move through three planes. This joint has a loose joint capsule which is surrounded by large and strong muscles; however, this stable joint allows for the range of motion required for daily activities such as walking, sitting and squatting (Middleditch & Oliver 2005). The hip joint plays a fundamental role in distribution of forces around the joint (Kim 1987). Cartilages line the bone end surfaces to allow the joint to rotate and move smoothly and freely in all planes of movement and to decrease the friction process. The cartilage lines the acetabulum surface, which pads the bones during weight bearing processes.

2.1.7 Hip ligaments and capsules

The complex system of ligaments that link the femur to the pelvis is vital for stability and restricts hip movement outside of its normal planes. While the ball and deep socket formation naturally provides more stability for the hip joint, the ligamentous capsule certainly contributes substantially. The joint capsules are strong and formed by the linkage of three separate units. The ligament with articulated between iliac and former called iliofemoral ligament which located anteriorly to the hip taking ‘Y’ shape (Byrne, 2010). It extends, in a spiralling form, from the ilium to the intertrochanteric line and it is taught in extension and relaxed in flexion, keeping the pelvis from tilting posteriorly in an upright stance and limiting adduction of the extended lower limb (Byrne, 2010). Inferior and posterior to the iliofemoral ligament, and blending into its medial edge, the pubofemoral ligament contributes to the strength of the anteroinferior portion of the capsule and this is perhaps the weakest of the four ligaments (Byrne, 2010). Posteriorly, the ischiofemoral ligament completes the main ligamentous constraints – from its ischial attachment origin medially its insertion laterally on superolateral aspect of the femoral neck, medial to the base of the greater trochanter (Byrne et al. 2010).

2.1.8 Hip muscles

Hip joint muscles have the responsibility of acting simultaneously to empower the hip to move in multiple directions, and to stabilise the lower extremity during weight-bearing activities, both when standing or walking. The major flexor of the hip joint is iliopsoas which comprises three muscles: psoas major, psoas minor and iliacus. The psoas major muscle extends from the twelfth vertebra of thoracic to the fifth vertebra of lumbar spine vertebral bodies and insets into the lesser trochanter (Schuenke et al 2006). Iliopsoas is the most powerful hip flexor, and is supported by sartorius, rectus femoris and tensor fascia latae (TFL) (Schuenke et al 2006). The gluteus maximus muscle is the most powerful muscle in the hip extensor. It contributes to abduction and adduction by its upper and lower fibres. The gluteus medius and minimus muscles contribute to hip abduction while piriformis contributes to hip external rotation and extension. From superior to inferior, these contain the superior gemelli, obturator internus, inferior gemelli, and quadratus femoris. Adductor longus contributes to hip flexion up to 70° (Schuenke et al 2006). Adductor brevis arises from the inferior pubic ramus and inserts proximal to adductor longus into the proximal third of the linea aspera. Adductor magnus arises from the inferior pubic rami, ischial ramus and ischial tuberosity. It inserts distally into the medial lip of the linea aspera but also has a more tendinous insertion into the medial condyle of the femur. It also contributes to extension and external rotation. Adductor minimus runs from the inferior pubic ramus into the medial lip of the linea aspera also contributing to external rotation. Gracilis is the only adductor that inserts distal to the knee joint. As previously stated, the muscles of the hip joint can contribute to movement in several different planes depending on the position of the hip, which is caused by a change in the relationship between a muscle’s line of action and the hip’s axis of rotation. For example, the gluteus medius and minimus act as abductors when the hip is extended and as internal rotators when the hip is flexed. The adductor longus acts as a flexor at 50° of hip flexion, but as an extensor at 70°. In addition to providing stability and motion for the hip, muscles act to prevent undue bending stresses on the femur. When the femoral shaft undergoes a vertical load, the lateral and medial sides of the bone experience tensile and compressive stresses respectively (Schuenke et al 2006). To resist these potentially harmful stresses, as might occur in the case of an elderly person whose bones have become osteoporotic and susceptible to tensile stress fractures, the TFL acts as a lateral tensioning band (Schuenke et al 2006).

2.2 Contribution of multi-spinal regions

In physical therapy departments, rehabilitation centers and orthopedic clinics, the spinal flexibility represents the functional performance of the spinal motion (Hsu et al. 2008). Spinal flexibility is considered as an important part of preoperative assessment and postoperative functional assessment (McGregor & Hughes 2004). Investigation the ROM of multi-spinal regions produces the potential to expand our perception regard the severity of spinal disorders, for instance, development of ankylosing spondylitis and the surgical influence on multiple-level discectomy or laminectomy (Hsu et al. 2008). Wong et al (2007) reported that the measurement of human spinal movement and posture is a very important part of research in the bioengineering and rehabilitation fields. A range of systems have been used to measure spinal range of motion in the cervical (Theobald et al., 2012; Tsang et al. 2013), thoracic (Hsu et al., 2008) and lumbar spine (Shum et al., 2010b; Hsu et al., 2008; Williams et al., 2013). Thoracic motions are believed to be relatively small motions, particularly at flexion/extension (AMA, 2000). A number of studies (Edmondston et al., 2012; Edmondston et al., 2011; Edmondston et al., 2007; Willems et al., 1996; Hsu et al., 2008; Mannion et al., 2004) have conducted the normal ROM of thoracic. Edmondston et al. (2012), Edmondston et al. (2011) and Edmondson et al. (2007) have examined the thoracic region separately from the mechanical interaction between thoracic motion and superior and inferior regions (i.e. cervical spine and lumbar spine). Contrarily, Hsu et al. (2008) and Mannion et al. (2004) have demonstrated both thoracic relative to lumbar motion. To the best of the author's knowledge, only one study other than Willems et al. (1996) measured the thoracic regions at different levels. The application of these methods to the spine often involves the use of two sensors, creating a hypothetical single 'joint' of interest. An example here would be to place one sensor at S1 and another at L1, thereby considering the lumbar spine as 'one region' (StamosPapastamos et al., 2011; Burnett et al., 1998; Lee et al., 2003) or on C7 and another at T12, thereby considering the thoracic spine as 'one region' (Hsu et al., 2008). The inherent limitation is that movement behaviour between the two sensors is not known (Williams et al., 2010). The literature, which measured the lumbar range of motion, typically applies two sensors or markers, one at each end of the lumbar spine. This includes technologies relying on electromagnetics (Shum et al., 2005; Shum et al., 2007), inertial sensors (Ha et al., 2013; Williams et al., 2013) and fibre-optics (Williams et al., 2010). Calculating the resultant angle between these two sensors provides an estimate of lumbar range of motion, with the lumbar spine measured as a single 'joint'.

The lumbar spine, however, consists of many regions or 'joints' (L1-S1) and thus examining this single joint region may result in missing information about regional lumbar spine movement behaviour. Dieën et al. (1996) have reported that changes to movement behaviour will not be obtained by any technique used, which measures the spine as a rigid region. Clinical decisions of abnormal spinal movement must rely on our basic knowledge of the natural dynamics of the spine. Authors have, however, commented that it is not sufficient to consider thoracic (Willems et al., 1996) or lumbar spine (Parkinson et al. 2013) as a single, rigid body when evaluating overall spinal movements. Therefore, practitioners must take into account the regional variations when making decisions on abnormality and when trying to base therapeutic movement protocols on theoretically derived patterns (Willems et al., 1996). The finding that the UL and LL are functionally independent is important for clinical practice. The majority of LBP has its origin in the lower lumbar spine but clinical assessment models as yet do not strongly encourage a focus on this region. Clinicians often observe the whole lumber spine during postural and movement assessments. The findings of this project could support the previous suggestion that, measuring lumbar spine as a single region may be an oversimplification. Regional management based on common restoration of movement or postural correction is unlikely to have a profound and targeted effect on the LL spine. Therefore, clinicians should concentrate on the movement and loading behaviour of the specific region associated with pain and dysfunction when faced with an LBP patient.

Clinical interest in sagittal kinematics has generally focused on flexion and extension range of motion between the twelfth thoracic (T12) and the first sacral (S1) vertebrae (Burton et al., 1989). However, Farfan (1975) suggested that the biomechanical function of the upper part and lower part of the lumbar spine may differ. Hilton et al. (1979) found age-related differences in regional motion in the UL (T12-L4) and LL (L4-S1) when they measured lumbar sagittal motion in vitro. Recognising that the lumbar spinal regions may differ in terms of functional support, the perception of the lumbar spinal regions providing the same contribution may not accurately reflect the pain and function in this region (Mitchell et al., 2008). Whilst previous authors have suggested that the UL and LL spine display differences in their kinematic behaviour as seen in Williams et al. (2013), traditional single 'joint' regions fail to identify such subtleties and, therefore, may oversimplify the description of movement. However, it is difficult to visualise this with the naked eye. The three main regions of the spine (i.e. cervical, thoracic spine and lumbar spine) are often assessed separately or the thoracolumbar spine is assessed as one region. Clinically, to deliver a more accurate assessment, clinicians should instead assess more than one anatomical region. Therefore, measurement of the spine to observe small details in multi-spinal regions movement is required to understand motion sharing within the spine (Gill et al., 2007; Parkinson et al., 2013). Investigation the ROM of multi-spinal regions could expanded our perception regard the severity of spinal problems which guide to select the appropriate protocol for treatment and follow up.

2.3 Lumbar-hip biomechanics

Biomechanical principles present a useful indication of the mechanisms of injury, movement behaviour and treatment programmes. The purpose of biomechanics research is to understand the very complex structure of the human body while experimental studies of biomechanics are used to demonstrate the mechanical properties of biological materials (Middleditch & Oliver 2005). The biomechanics of the musculoskeletal system requires considerable knowledge of different fields that may include neurophysiology, physiology and biomechanics (Middleditch & Oliver 2005). The aspect of lumbar and hip biomechanics in this research (i.e. the experimental studies) is kinematics. The kinematics is the branch of biomechanics that enables the object's motion regardless of the forces involved (White 3rd & Panjabi 1978). The lumbar-hip complex range of motion and velocity are considered aspects of kinematics. Range of motion is an angle through which a joint transfers from an anatomical position to the extreme limit of regional motion in one direction (Lee 2006) while velocity is the rate of change when a joint transfers through a particular direction (Freeman and Lawlis 2004).

When specialists (i.e. physiotherapists or chiropractors) discuss spinal movement, the spinal functional unit is usually presented in such fields. The segmental motion is a spinal function component and the smallest functional part, which represents the biomechanical behaviour of the entire spinal column (Wang 2012). The segmental motion consists of two vertebrae, intervertebral disc (IVD), facet joints, capsules and ligaments (Middleditch & Oliver 2005). The movement between two vertebrae is limited, however, when movement between numerous vertebrae is combined, greater movement will be evident (Cramer & Darby 2013). The IVDs contribute to the limitation of motion, which occurs between two vertebrae; therefore, thicker IVDs of cervical and lumbar spine help to increase the range of motion in these regions (Cramer & Darby 2013). In addition, the articular facet's form and orientation control the movements that can occur between two adjacent segments and decrease the magnitude of the movement that may occur between segments (Middleditch & Oliver 2005). Spinal movement is also restrained by bony stops and ligamentous control (Louis 1985).

The primary movement of the spine comprises three planes: the frontal plane, the sagittal plane and the transverse plane. Spinal flexion/extension moving in sagittal plane, lateral flexion to right and to left moving in the frontal plane and spinal rotation to right and to left moving in the transverse plane. There are some structures associated with each type of spinal movement such as flexion, extension and later flexion (Cramer & Darby 2013). In flexion, the anterior longitudinal ligament is relaxed while the posterior longitudinal ligament is tough when the anterior portions of discs and vertebrae are compressed. Consequently, the distance between laminae becomes wider, and the inferior articular processes glide upward on the superior articular processes of vertebrae (Cramer & Darby

2013) while the reverse process occurs during extension movement. Lumbar and cervical regions allow for more flexion than the thoracic region. The limited motion of the thoracic spine in extension is due to thinner discs and the structure of the skeleton as well as the muscular structure of the thoracic (Williams et al., 1989). The lateral flexion motion compressed on sides of IVDs is greater at the cervical region, followed by the lumbar region and finally the thoracic regions (White & Panjabi 1990).

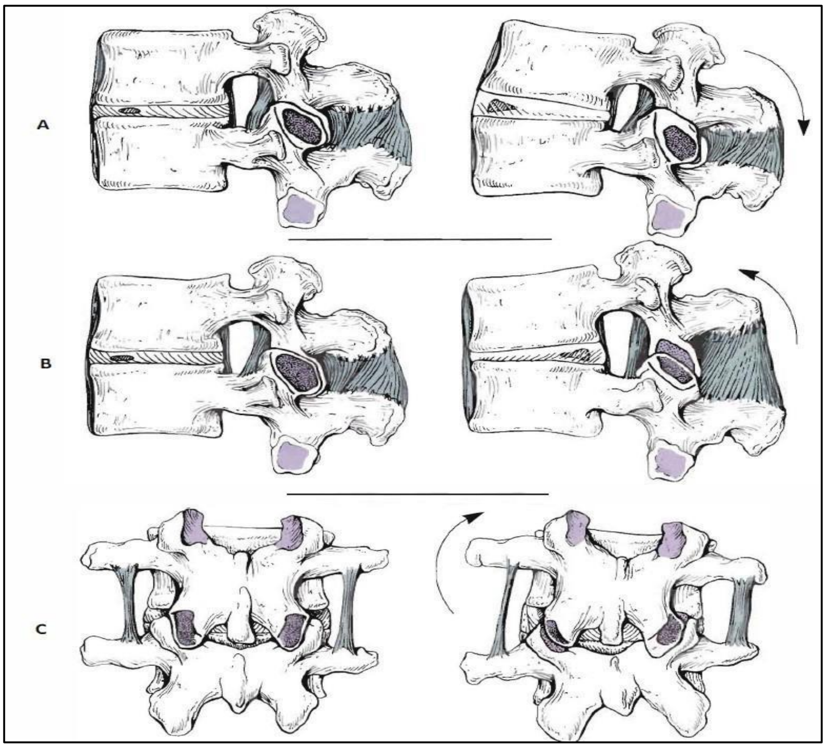


Figure 2.2.1: The primary planes of spine are frontal plane, sagittal plane and transverse plane (Gramer & Darby 2013)

A considerable number of studies have focused on the spinal flexion/extension movement, disregarding the relationship between lumbar and hip (Van Herp et al. 2000; Pearcy 1985; Lee and Wong 2002; Lee et al. 2003; Hindle et al. 1990; Russell et al. 1993; Ha et al. 2013). These studies have addressed both flexion and extension when they measured the cardinal movements; however, Milosavljevic et al. (2008) have examined spinal extension movement only.

These studies have measured the range of motion of lumbar spine flexion and extension that showed differences in their findings possibly due to individual factors such as differences in gender, body weight, body structure, job and other influential factors (Ha et al. 2013). They have used lumbar spine as a 'single' region, while it is known clinically that an individual who suffers from lower back pain reports more pain in the lower lumbar spinal regions than the upper lumbar regions (Biering-Sørensen 1983; Beattie et al. 2000). This assert is also supported by the fact that the lower lumbar segments are consistently more susceptible to degeneration than upper lumbar regions (Quack et al. 2007; Twomey & Taylor 1987). This degeneration is thought to be due to higher mechanical stress through the lower lumbar regions (Adams & Hutton 1983).

The dominant functional tasks such as flexion, extension, object lifting from floor and transiting from stand-to-sit or sit-to-stand are associated with spinal disorders and functional impairment. Spinal mobility impairment leads to various forms of functional disabilities (Cox et al. 2000), which may have a serious impact on an individual's quality of life. However, such activities are also known to be affected by the presence of disorders of the lumbar-hip complex. In many daily activities, spine and hip kinematics are closely coordinated (Mayer et al. 1984; Strand and Wie 1999). This suggests that disorders of the lumbar-hip complex may affect functional tasks as well as the cardinal movements often employed in the clinic. Sit-to-stand and stand-to-sit activities are regular daily functions (Lomaglio and Eng 2005). The most important task that utilizes lumbar and hip motion is object lifting from the floor, which is a common daily activity particularly amongst those working in jobs involving physical labour (Shum et al. 2005). Therefore, authors have previously studied biomechanical functions of lumbar spine relative to the hip using various measurements in different tasks such as movement from flexion to upright standing, from upright standing to extension or flexion, object lifting, sit-to-stand or stand-to-sit (McGill 1997; Dempsey 1998. There is an associated effect on the lumbar-hip complex constraining the range of motion and movement behaviours due to lower back pain when performing daily life activities (Shum et al. 2007a).

As the assessment of an individual with spinal problems usually involves the completion of movements in the cardinal planes, for example flexion/extension, the relationship between flexion profile task and other cardinal functions such as lifting, stand-to-sit and sit-to-stand must be established. The lumbar-hip complex is an anatomical region, which influences the body structures and is associated directly with both the upper body and lower extremities (Bruno 2014). The researchers claimed that the evaluation of spinal and hip behaviour during functional tasks is very important particular flexion movement considered as essential protocol in spinal examining to identify lumbar spine problems. The extension movement also considered an indicator to differentiate between healthy subjects and lower back pain sufferers as it is a fundamental component associates with several functional activities (Milosavljevic et al. 2007).

Forward bending and objects lifting activities are related to lumbar disc and ligament injuries and cadaveric experiments propose that this harm is mostly attributable to a great bending moment influencing the osteoligamentous spine (Dolan & Adams 1993).

It has been recognised that work-related lifting, (Marras et al. 1993) as well as mechanical factors related to lifting, are risk factors for lower back pain (Waters et al. 1993; Ferguson & Marras 1997). Object lifting from the floor is a common daily activity especially amongst those working in jobs involving physical labour (Shum et al. 2005); therefore authors have focused on the effect on the lumbar spine caused by object lifting (Larivière et al. 2000; Kingma et al. 2001; Marras et al. 2001; Shum et al. 2007).

Sit-to-stand and stand-to-sit activities are regular daily activities (Lomaglio and Eng

2005), which are performed on average sixty times per day by working people (Dall and Kerr 2010). Authors often try to detect ideal sit-to-stand biomechanics (Janssen et al. 2002), to rehabilitate this functional task. The known ideal biomechanics for this task will emerge from understanding typical performance (Shum et al. 2005a; Fotoohabadi et al. 2010; Kuo et al. 2010).

Stand-to-sit and sit-to-stand movements are not simple tasks because they are the most mechanical and muscularly demanding tasks carried out during daily activities (Aissaoui & Dansereau 1999; Dubost et al. 2005; Faria et al. 2010). However, the ideal coordination, balance, perfect movement and optimal strength, and muscle power are necessary for good stand-to-sit and sit cycle accomplishment (Cadore & Izquierdo 2013). In particular, the geriatric population faces difficulties when performing such activities (Alexander et al. 1991). The period of time which is spent by an old person moving to sit appears to increase (Riley et al. 1991; Rodosky et al. 1989) as does the level of difficulty faced when standing up independently due to diminished functions and movement in daily living activities, which increases the risk of falling (Nevitt et al. 1989; Tinetti et al. 1995). It was recommended in rehabilitation strategies to extend the spine to achieve successful performance of sit-to-stand; however, kinematic results indicate lumbar flexion usually occurs during this task (Schenkman et al. 1990; Tully et al. 2005).

The velocity of the lumbar spine regions and the hip is also necessary in measuring the quality of kinematics. The motion magnitude and velocities, as well as the coordination between the lumbar spine and hips, are influenced considerably by the presence of back pain, particularly in subjects with lower back pain with a positive straight leg raise (SLR) (Shum et al. 2010). There is also some evidence to suggest that lower back pain influences the timing of the spine and hip during flexion movement (Wong and Lee 2004). Timing here means the hip movement of lower back pain sufferers occurred before lumbar spine during extension task. Experimental data related to the timing and combination of the spine and hip in healthy subjects performing less limited movement tasks provide valuable information regarding coordination of the trunk (Thomas & Gibson 2007). In contrast, coordination of the lumbar spine and hip joint movements during flexion and extension was assessed in healthy subjects and those with lower back pain found no timing differences between them. Granata and Sanford (2000) have stated that spine and hip kinematics happen simultaneously during flexion and extension movement when a subject is carrying weight. Contrarily, Nelson et al. (1995) found the spine and hip movement to occur simultaneously in the trunk flexion movement; however this hip movement occurs before spine movement when returning to an upright position.

Researchers have previously been interested in studying the relationship between lumbar spine and hip movement, as well as also interested in the relationship between lumbar and pelvis. Addressing the relationship findings between lumbar spine and pelvis will open the gateway to imagining the movement behaviour of lumbar-hip regions.

The relationship between lumbar-pelvis complex and hip movements suggests that regional movement of the pelvis is related to its function in hip flexion as the pelvis moves around the thighs during forward flexion tasks (Johnson et al. 2010). Johnson et al. claimed that there is greater variability of range of motion in the lumbar and pelvis joint compared to the other spinal joints. The relationship between lumbar and pelvis during flexion and extension was usually examined from an upright position with the knee extended (McClure et al. 1997; Pal et al. 2007; Kuo et al. 2010; Kanayama et al. 1996; Tully et al. 2002; Esola et al. 1996; Kim et al. 2013). These studies have revealed diversity in results, but a considerable number of them have indicated that the lumbar spine moves first followed by the pelvis during flexion tasks and vice versa at extension. The first half of flexion movement occurred at the lumbar spine (45º) while the pelvis remained relatively fixed while the second half was accomplished by forward movement of the pelvis (Cailliet 1981).

Escalating the belief that the lumbar spine regions possess different contributions, regional motion profile of the lumbar spine and hip and the interaction with the lumbar spine are considered important (Lee and Wong 2002; Sahrmann 2002; O'Sullivan 2005).

2.3.1 The lumbar-hip complex movement during flexion/extension

The assessment of flexion and extension movements of spine and hip is considered an essential technique for doctors and physiotherapists to identify spinal disorders (Esola et al., 1996; McClure et al., 1997; Porter and Wilkinson, 1997). The best way to understand the relationship between the lumbar spine and hip is to examine this relationship during spinal flexion and extension (Johnson et al., 2010).

An evaluation of the relationship of the lumbar spine relative to hip during flexion and extension tasks was conducted in a number of studies (Shum et al. (2007a), Wong and Lee (2004), Lee and Wong (2002), Shum et al. (2005a), Paquet et al. (1994)), which asserted that the contribution of the lumbar spine to forward bending was less in subjects with LBP. However, other studies reported that the contribution of the lumbar spine increased in subjects with back pain and in healthy subjects with a history of back pain (Esola et al., 1996; Porter and Wilkinson, 1997).

Paquet et al. (1994) also investigated the lumbar spine and hip movement involving muscle activity measurement in subjects with and without LBP, and found that the subjects with lower back pain had less movement at lumbar and hip than healthy subjects. Returning from full flexion to extension, the hip moved first (25% of the movement cycle) and the lumbar spine was predominant for the rest of the movement cycle (75% of the last movement cycle). Mayer et al. (1984) examined healthy subjects and found the lumbar flexion range of motion to be about 55º and hip flexion to be 66º. Porter and Wilkinson (1997) found that the relative motion of the lumbar spine and hip in standing to full flexion differed between healthy subjects and chronic LBP sufferers who were examined bending forward from an upright standing position to full flexion. In particular, hip movement was shown to be slower. Furthermore, Porter and Wilkinson suggested that their findings reflected the importance of evaluating the lumbar and hip flexion movement especially with regard to chronic LBP patients, in order to ascertain whether there was abnormal movement.

Cailliet suggested that forward function behaviour was controlled neurophysiologically while his results supported the findings reported by Farfan (1975). The first 50-60º degrees of spinal flexion occurs in the lumbar spine, particularly in the lower part of the lumbar, with the pelvis then tilting forward allowing for more spinal flexion (Farfan, 1975). These studies implied that understanding the spine-hip relationship during flexion and extension is an important indicator that provides more information about lumbar spine impairment.

The ratio of spine-to-hip has also been studied, when rising up from full flexion to full extension with the patient either carrying an object or not (Nelson et al., 1995; McClure et al., 1997; Granata and Sanford, 2000). The researchers found a higher contribution from the spine than from the hips at the early stages of movement, while the contribution of the hips was more than that of the spine at the final stages of the movement (Paquet et al., 1994; Esola et al., 1996; Porter and Wilkinson, 1997; Lee and Wong, 2002; Wong and Lee, 2004). Ratio alterations are not only a sign of lumbar and hip complex disorders but are also an indicator that the bending and compressive stresses of lumbar spine are being affected in healthy subjects (Tafazzol et al., 2014; Dolan and Adams, 1993).

While the findings from these studies provided some information on the spine and hip relationship, there were differences in spine and hip movement patterns, which may be due to the difference in starting time at these joints (Thomas and Gibson, 2007). The variance in these studies' findings might also be due to differing techniques, system measurement, age of subjects, starting position and location of sensors/markers. In addition, measuring lumbar spine as a single region could also be a factor in the variation in results as well as differences in LBP history or physical appearance of the subjects being studied (Wong and Lee, 2004).

Table 1: Comparing previous literature that examined the whole lumbar spine, upper and lower lumbar spines with relative to hip movement or without during flexion/extension movement

Study	**WLS**	**ULS**	**LLS**	**Hip**
Shum et al. (2007a)	√			√
Shum et al. (2005a)	√			√
Wong and Lee (2004)	√			√
Lee and Wong (2002)	√			√
Paquet et al. (1994)	√			√
Mayer et al. (1984)	√			√
Nelson et al., (1995)	√			√
McClure et al., (1997)	√			√
Granata and Sanford 2000)	√			√
Esola et al. (1996)	√			√
Porter and Wilkinson (1997)	√			√
Parkinson et al. (2013)		√	√	
Mitchell et al. (2008)		√	√	
Leardini et al. (2011)		√	√	
Williams et al. (2012)	√		√	
Williams et al. (2013)	√		√	

- WLS, whole lumbar spine; ULS, upper lumbar spine; LLS, lower lumbar spine

Greater motion contribution from the lower lumbar spine may help to explain an increased prevalence of LBP in the LL than the UL spine (Biering-Sørensen, 1983; Beattie et al., 2000). Table 1 shows the need for establishing more detailed information about UL and LL contribution relative to hip during flexion and extension, which could provide new knowledge for clinicians who may then be able to use a more suitable form of assessment and treatment. Table 1 shows that the authors have adopted multi-regional lumbar spine regions with no regard for hip movement across clinical populations (Williams et al., 2012; Williams et al., 2013) and healthy subjects (Leardini et al., 2011; Parkinson et al., 2013), revealing differences in regional contribution. Even though previous studies have suggested that the UL and LL spine may be functionally different, a greater understanding of the interaction between these regions is required (O'Sullivan, 2005). Therefore, the behaviours of spinal movement and the differences between the regions relative to hip require further study. Obtaining such information could expand the knowledge of physiotherapists and improve their clinical assessment of LBP sufferers.

2.3.2 The lumbar-hip complex movement during lifting movement

The kinematics of the lumbar spine and hip when moving up from full flexion to upright standing or bending to lift objects have been investigated by various authors (Porter and Wilkinson 1997; Paquet et al. 1994; Nelson et al. 1995; McClure et al. 1997; Dolan and Adams 1993; Lee et al. 2001; Wong and Lee 2004; Shum et al. 2007). Findings by Shum et al (2007) confirmed differences in ratio between lumbar and hip in those with and without back pain as there was more movement shown in the hip than the lumbar spine during a picking-up activity by about 50% in healthy subjects. While the lumbar contribution appeared to reduce in subjects with lower back pain. Their study protocol was designed to investigate the relationship between kinematics of lumbar spine and hip joint as well as coordination when picking up an object from a sitting position. They reported that the ratio of lumbar spine relative to hip was about 0.53 in healthy subjects, while this reduced in subjects with lower back pain to 0.45. Thomas et al. (1998) assessed the relationship between the spine and hip in free full body reaching tasks and discovered that the spine–hip ratio was 1:2.

McClure et al. (1997) determined the movement pattern of lumbar spine and hip on 12 healthy subjects and 12 subjects with lower back pain from full flexion to upright position. They found lumbar spine movement behaviour was different from hip movement behaviour while earlier movement at lumbar spine was shown in subjects with lower back pain, particularly in the initial 25% of the movement. McClure et al. reported that the hip contribution at the beginning of the extension (rising to upright standing) was greater than that of the lumbar; however, lumbar contribution increased in the middle of the movement and showed the primary source of movement during the final stage. However, healthy subjects with poor mobility in sagittal plane displayed an increase in the magnitude of bending stress on the lumbar spine during forward bending and lifting activities (Dolan and Adams 1993). Significantly, this may increase the risk of injury to the intervertebral discs and ligaments.

Lumbar spine kinematics during object lifting task has recently been studied as two regions. Considering, the lumbar spine as more than one region has been advised by Gill et al. (2007), who collected data from the mid thoracic spine, lower thoracic/upper lumbar spine, mid lumbar spine, and the lower lumbar spine at lift onset. They found no variation in lower lumbar spine posture at a lifting start, regardless of the lifting technique used, or the distance between subjects' feet and the object on the floor. However, Gill et al. reported that movement variation during lifting tasks occurred in the upper lumbar spine and mid thoracic spine, but not in the lower lumbar spine. They claimed that tensile strain on tissues in the lower lumbar region that might carry a risk of injuries in lifting was not affected by the lifting style or horizontal lift distance when lifting from floor level. Mitchell et al. (2008) have reported a lack of correlation between upper and lower lumbar regions which supported the findings by Gill et al. that the contribution of the lower lumbar spine was greater than the upper lumbar spine. Importantly, these differences between the upper and lower lumbar spine regions were only evident when the lumbar spine was measured as two separate regions (Dankaerts et al. 2006). As object lifting from the floor is a common daily activity, which causes lower back pain, there is still a need to obtain more information related to movement behaviour and the relationship between hip and lumbar spine as two regions.

A lack of information about movement behaviour of lumbar regions relative to hip in the healthy population during object lifting has encouraged the study of the lumbar spine as two regions in this task. Furthermore, no previous authors have demonstrated the kinematics of the regional lumbar regions i.e. upper and lower lumbar spine relative to hip movement during object lifting. Therefore, two regions of the lumbar spine will produce more information regarding movement behaviour during object lifting with clinicians able to adapt new procedures for their assessment that may improve health conditions of sufferers.

2.3.3 The lumbar-hip complex movement during stand-to-sit and sit-to-stand

Sit-to-stand movement has been examined in healthy subjects (Janssen et al. 2002; Fotoohabadi et al. 2010; Kuo et al. 2010; Johnson et al. 2010; Leardini et al. 2011; Parkinson et al. 2013) and in lower back pain sufferers (Shum et al. 2005b; Hsieh & Twomey 2010; Boonstra et al. 2011; Shum et al. 2005a). These studies investigated the movement within the spine during sit-to-stand and conducted different information about range of motion of lumbar spine. However, previous studies findings diversity may due to different positions of markers or sensors, which lead to conflict data. Spinal assessments and effects of treatment data are not available as there is no standard spine region (Parkinson et al. 2013). Two studies, however, which examined two tasks in patients (Shum et al. 2005a; Shum et al. 2007a) reported that the movement of lumbar spine and hips considerably decreased and that subjects with lower back pain moved in different ways in order to compensate for the limited movement at lumbar spine and hips. Coordination between lumbar spine and hip was significantly altered and muscle contractions at the lumbar spine and hip during stand-to-sit and sit-to-stand were found to be reduced in subjects with lower back pain.

Even though clinicians typically separate the lumbar during assessment (Dankaerts et al. 2006), the majority of the literature has considered movement of the lumbar region as a singular entity. Given the dispute related to the ideal spinal kinematics and with the intention to provide full understanding of lumbar spine kinematics, breaking down the lumbar spine into two regions (i.e. upper and lower spine) to measure the movements in stand-to-sit or sit-to-stand in healthy subjects is necessary (Mitchell et al. 2008). In addition, considering the spine as a single rigid body may not be sufficient to reflect spinal kinematics (Parkinson et al. 2013). Leardini et al. (2011) have investigated upper and lower lumbar regions but only in small samples (n = 10 subjects). Even so, they found considerable differences between upper and lumbar spine regions and suggested that two functionally independent lumbar regions must be confirmed in a large sample. To confirm the results of Leardini et al. (2011), Parkinson et al. (2013) determined the difference between lumbar spine regions (whole lumber, lower and upper spine) moving from sitto-stand in 29 asymptomatic subjects and found that the lower lumbar region moved about two-thirds more than upper lumbar region during sit-to-stand.

Rising from sit-to-stand and vice versa are fundamental and functionally important activities. Previous studies have revealed that not only are there considerable decreases in lumbar spine and hip motion during sit-to-stand and stand-to-sit but coordination also altered within the lumbar spine-hip joint complex (Shum et al. 2005a). Therefore, authors have focused on kinematics of lumbar spine either as one single region or separated into two regions (upper and lower lumbar spine) regardless of hip motion.

2.3.4 Lumbar and hip velocity

Visualising kinematics control is important because previous studies have exposed that LBP is related to altered control (Van Dillen et al., 2009; Hodgs et al., 2009; Xu et al., 2010). Some studies (Shum et al., 2005; Williams et al. 2013) have examined the ROM-angular velocity relationship which is known as the spatial relationship (ROM-angular velocity plot). The plotting of ROM-angular velocity provides a useful clinical picture of spine dynamics, in which the emphasis is on representing the kinematic control of the system (Li et al., 1999). This clinical approach can be easily accomplished with the use of inertial sensors (Saber-Shiekh; Theobald et al., 2012). Lumbar spine and hip movement has been studied using the kinematic parameters of angular displacement, velocity and acceleration (Esola et al. 1996; McClure et al. 1997; Lee and Wong 2002; Wong and Lee 2004; Pal et al. 2007; Shum et al. 2005b; Williams et al. 2013). Impairment of one of these variables is considered to be an important sign indicative of spinal problems (Marras et al. 1995; Marras and Wongsam 1986; McClure et al. 1997; Williams et al. 2013). For instance, the findings of these studies either on symptomatic or asymptomatic subjects indicate that velocity when the spine is extended is significantly decreased in subjects with lower back pain and that this reduction is greater than that observed in flexion (Marras & Wongsam 1986; Marras et al. 1995). Even though the velocity of extension is considered a useful measure to serve as an indicator of spinal musculoskeletal condition (Marras & Wongsam 1986) there are still some confounding findings on the main variable normally used when classifying subjects with lower back pain on the basis of timeindexed kinematic features. Authors have reported significant reductions in the magnitude of lumbar spine movement in all directions. For instance, reducing hip flexion and changing hip and lumbar spine kinematics in subjects with lower back pain were found at the execution of spine movements (Wong and Lee 2004) and lumbar spine movement during extension (Gombatto et al. 2007; Van Dillen et al. 2007). The clinical consideration has been to test the kinematics of changes in spinal shape, which revealed a greater difference between subjects with lower back pain and healthy subjects than measuring; for example, range of motion only (Consmüller et al. 2012). Marras & Wongsam (1986) investigated the importance of dynamics during functional activities on 16 lower back pain patients and 18 asymptomatic subjects. Even though they found a decrease of 10% in the flexion range of motion in patients suffering from lower back pain compared to asymptomatic subjects, the substantial decrease of 50% in angular velocity was a considerably clearer biomarker for lower back pain sufferers (Consmüller et al. 2012). Furthermore, the angular velocity of patients during extension movement reduced by more than 90%.

Previous studies that have investigated the extension kinematics have focused either on thoracic spine examination relative to the hip (McGregor et al. 1995) or on the relationship between lumbar spine and pelvic movement, from the fully flexed to the upright position (McClure et al. 1997; Paquet et al. 1994). McGregor et al. (1995) examined 20 subjects with lower back pain and 20 healthy subjects using the CA-6000 devices, concluding that the spine velocity during flexion movement in the sagittal plane was clear functional proof of disorders.

Milosavljevic et al. (2008), who found that the lumbar spine when starting the movement extended backward and returned to the standing position significantly earlier than the hip, examined the lumbar spine and hip kinematics from an upright position to full extension.

The findings, which were conducted on healthy subjects, reflected the tendency for the lumbar spine to dominate over the hip during extension to the end ROM of spinal backward movement in terms of the magnitude and velocity of the movement.

Two studies (Shum et al. 2005b; Shum et al. 2005a) have examined the relationship of range of motion-angular velocity. This association, known as the spatial relationship, can be seen through a range of motion-velocity (Williams et al. 2013).

Earlier studies have only evaluated the velocity across the lumbar movement as a single region rather than analysing subunits of movement. Identifying velocity in multi-spinal regions relative to hip will provide more detailed information that may help to design specific therapeutic exercise interventions for patients with lower back pain. As movement velocity in subjects with lower back pain reduces to protect and limit movement behaviour against extreme loading and parallel pain, it appears beneficial to improve clinicians' knowledge about multi-regions' velocity with regard to the upper and lower lumbar spine and hip joint. The importance of velocity is high in monitoring the movement behaviours, which yields information as digital values as well as curves of movement behaviour capturing over time and comparing between regions' values of velocities. Angle-angle plots will provide a description of where the range of motion of each region are against one another, thereby revealing further insights into kinematic behaviour.

2.4 Spinal motion measurement techniques

Assessment of spinal kinematics through observation is a fundamental part of the clinical cognitive process. The observation can identify key features that would affect the direction of treatment approach. Unfortunately, the physiotherapist's eye may have difficulty in detecting the precise degree of spinal and hip kinematics and the contribution of each particular region at the spine, which then makes it impossible to know the relationship between these regions. Hence, the researchers used various measure systems that are able to demonstrate the changes related to functions impairment or monitoring the improvement with treatment and rehabilitation programmes (Lebel et al. 2015). Traditionally, the mobility assessments are tested using self-reports of patients who answer a list of questions or by testing the performance of an examination that uses different clinical measurement systems. However, sophisticated systems using either invasive or non-invasive measurement methods usually measure spinal movement. Non-invasive techniques might be used in clinical applications or in more complex laboratory measurements. The precise three-dimensional (3D) systems that capture the dynamic movement in real-time require a complex setting and the current systems usually used to measure spinal movement are optic tracking, electromagnetic techniques, and goniometers and inclinometers.

2.4.1 Invasive measurement systems

Invasive systems used to measure spinal movement or shapes are planar and biplanar Xray (Pearcy 1985; Thoumie et al. 1998), Ultrasound-based coordinate measuring system (CMS), 70p (Zebris) system (Lee et al. 2006; Malmström et al. 2006) and magnetic resonance imaging (MRI) (Fujii et al. 2007). The MRI system uses magnetic fields and radio waves to create shaped images of the body part being examined. This system is used for various medical purposes such as diagnosis, and for monitoring the improvement in follow-up procedures without exposure to hazardous radiation. This system offers a precise measure of maximal spinal range of motion; however, this operation is very expensive and time-consuming. Furthermore, ultrasound and MRI are not adequate when measuring motion from an upright standing position, but X-ray techniques currently offer portability as it is used for regional level analysis (Gajdosik and Bohannon 1987).

Figure 2.4.1: Schematic representing the magnetic resonance imaging (MRI) (A), Computerized Tomography (CT scan) (B), Ultra-Sound (C).

The use of radiographs is the gold standard method which allows for clear visualisation of bony parts (Sprigle et al. 2002); however the hazards of radiation prohibit its extensive use in research trials (Perry et al. 2008). Furthermore, this option is not always available to practitioners in physical therapy clinics (Fitzgerald et al. 1983).

2.4.2 Non-invasive measurement systems

Non-invasive techniques are used in simple clinical methods or more complex laboratory systems. These simple techniques have been used widely in medical applications due to their portability; however, there are more complex constraints on laboratory applications.

2.4.2.1 Traditional measurement systems

Goniometers measure the angle between two arms of the device, lining up with body parts and bony landmarks, capable of motion in one plane (Figure 2.3.2, A). It is a useful clinical tool that is capable of evaluating the objective measurements to measure the improvement in rehabilitation routes. Another traditional measurement system is the inclinometers used for measuring angles of incline (i.e. tilt), moving up and down in pendulum movement, tracking the object's motion with respect to gravity (Figure 2.3.2, B). A new portable inclinometer product called Spine Mouse is used for quantifying the spinal shape and movement in the sagittal and frontal plane (Figure 2.3.2, C). This measurement system manually covers the spinal skin along the spinal column.

Measurement tape is a tool capable of measuring distance. This traditional tape measure was used to measure the distance between marks on spinal regions during spinal examination or fingers to floors are measured using measurement tape (Mellin 1989). Cervical range of motion device (CROM) is a group of inclinometers which are attached to a frame similar to glasses (one, in the frontal plane, to lateral inclination; another, in the sagittal plane, to flexion/extension; and the third, in the transverse plane, for rotation). Two of them are gravity-dependent (the frontal and sagittal) while the transverse one is magnetic dependent (Figure 2.3.2, E).

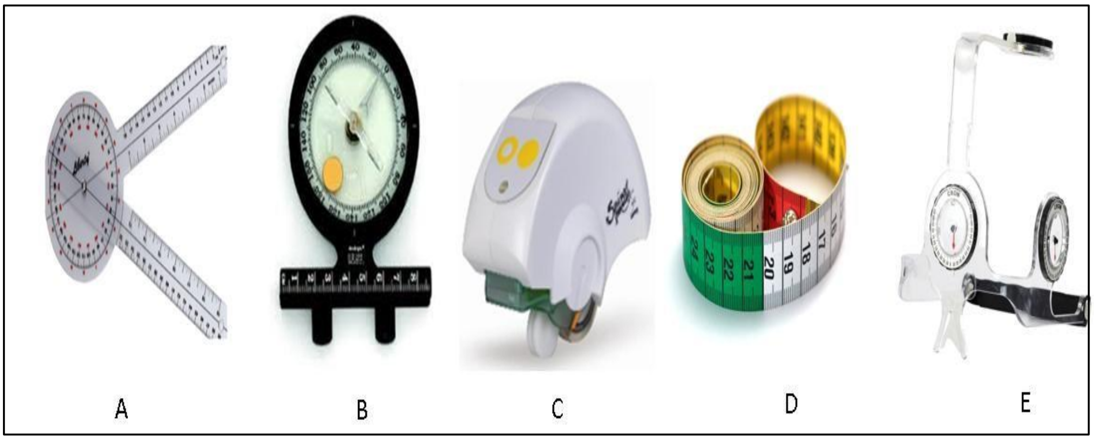


Figure 2.4.2: Schematic representing the Goniometer (A), Inclinometer (B), Spine mouse (C), Tape measurement (D) and Cervical Range of Motion (CROM) device (E).

Numerous traditional goniometers were used such as universal goniometer (Pearcy & Hindle 1989; Hindle et al. 1990; Herp et al. 2000; Sprigle et al. 2003; Yankai & Manosan 2009), the Spin-T goniometer (Agarwal et al. 2005a; Middleditch & Oliver 2005) and the electro-goniometer (Thoumie et al. 1998). The inclinometer devices measure tilting with respect to gravity and have also been used to measure spine movement and shapes (Youdas et al. 1991; Lynch-Caris et al. 2008), traditional inclinometer (Mayer et al. 1997; Mannion et al. 2004) and spinal mouse (Kellis et al. 2008; Mannion et al. 2004).

Goniometers and inclinometers are frequently used in clinics; however, they are limited to measuring a single joint. They are useful for measuring the angle between an initial and final position of a single joint or the spine, but are limited due to location slips and often act as a mechanical restriction. However, measuring multiple angles can be timeconsuming. Moreover, these tools are not valid for measuring the dynamic spine movement in real-time while movement behaviour across time is also lost (Williams et al. 2013). Meanwhile, the tape measurement is useless when a clinician's intent is to obtain angles from initial position to final position as it can be measure distance only (Williams et al. 2013).

2.4.2.2 Optical tracking systems

In the past three decades, a variety of tracking technologies, including optical tracking and electromagnetic tracking systems, have been developed for movement data, capturing and tracking a range of aspects such as entertainment, sports and medical applications (Kindratenko 2000). However, each tracking system has its own advantages and disadvantages compared to other systems along with the nature of the system and the applied areas (Kindratenko 2000). Optical tracking systems contain units of receivers including cameras and markers attached to the object. The system can measure the positions of the markers using geometry and image processing on the images acquired from stereoscopic cameras.

There are numerous optical tracking systems used for motion analysis such as vicon and Qualisys motion capture systems. The vicon motion capture system is an infrared marker tracking system that offers millimetre resolution of angular displacement in 3D. Qualisys motion capture system uses numerous high-speed cameras to capture the object's motion and it is precise and produces high-quality data for the observer in real-time. Optical motion analysis system (Edmondston et al. 2007), the digital optoelectric instrument (Sforza et al. 2002), 3space Isotrak system (Pearcy & Hindle 1989), motion qualisys system (Alenezi et al. 2014) and Vicon Motion Systems (Schache et al. 2002; Windolf et al. 2008) have been used for human motion tracking. Nevertheless, optical tracking systems are used routinely all over the world because of their ability to capture motion in realtime, something that would be difficult to achieve in clinical practice.

Such sophisticated technological systems make it possible to capture spinal kinematics during the performance of physical functions; however, laboratory setting requirements as well as high costs and time consumption for operation and processing make them unfeasible for clinical applications (O'Sullivan et al. 2009; Smith et al. 2010). An optical motion analysis system allows for very accurate recording of motion over time; however, the constrained field of view, in tandem with the required 'line-of-sight' and very specific illumination criteria, limits the use of such systems. Moreover, setting and calibration can be time-consuming (Lee and Wong 2002). The technique is also affected by obstruction problems, where the cameras cannot accurately identify a marker during the motion, due to lost position data. Instrument systems such as 3D capture of motion with optic or magnetic tracking systems are accurate but very expensive, and complex to configure and operate for clinicians (Zhou & Hu 2008). Accurate tracking of optical systems is usually limited to a specific environment which should have clear line of sight between various cameras and the reflected markers. Furthermore, the motion capture volume is usually controlled in space and the equipment (camera, transmitter and receiver) has to be calibrated in a specific environment to demonstrate its accuracy.

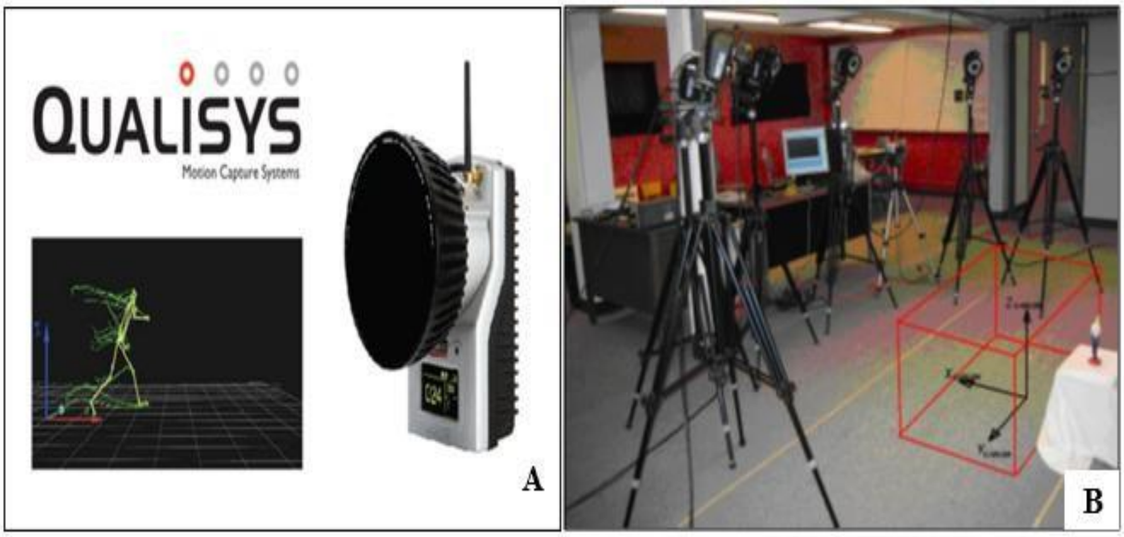


Figure 2.4.3: Schematic representing the optical motion system; A: Qualisys motion capture system which uses numerous high-speed cameras to capture the object's motion and it is precise and produces high-quality data for the observer in real-time and B: The vicon motion capture system which is an infrared marker tracking system that offers millimetre resolution of angular displacement in 3D.

The limitations of the setting, calibration, accessibility, excessive length of time required, constrained field of view and cost for these devices encourage companies to instead develop devices that are safe, inexpensive and portable. Usually, portable tools are used to assess spinal kinematics and it is crucial for these important measures to be considered by clinicians in order to support clinical decision-making (Consmüller et al. 2012).

2.4.2.3 Electro-magnetomtry

The last two decades have seen increased usage of electromagnetic tracking systems (EMTS) in medical applications. Generally, these systems consist of three components field generator, sensor unit and central control unit. The sensor unit and central control unit (field generator) which uses several coils to generate a position differing magnetic field that is utilised to create the coordinate space (Win 2010); the sensor unit which contains small coils attached to the body in which a current is generated via the magnetic field, therefore behaviour calculation of each coil can determine the position and orientation of the object (Win 2010). Using such techniques helps to demonstrate the position of sensors movement in space when the central control unit works to control the field generator and capture data from the sensor unit. One of the most important advantages of electromagnetic tracking systems is that electromagnetic fields do not rely on line of sight for operation. However, magnetic fields may interfere with the tracking units due to the presence of any electronic device that produces electromagnetic interference (Win 2010).

Figure 2.4.4: Electromagnetic Tracking Systems for Medical Applications:A. Polhemus Fastrak, B. Ascension microBIRD and C: NDI Aurora (Win 2010).

Electromagnetic tracking systems have been reported to be an advisable measure as they record with high accuracy in both biomechanics research (Mills et al. 2007) and in clinical practices (Lee et al. 2001; Jasiewicz et al. 2007; Mills et al. 2007). However, the possibility of errors remains high due to the presence of metal and these errors should be correct to avoid distortion of the magnetic field (Lebel et al. 2015). Correcting the metal distortion is both time-consuming and complex (Lee et al. 2003) Bull & McGregor 2000; Hsu et al. 2008).

2.4.2.4 Inertial sensing in human motion measurement

Accelerometry has a long history in human movement and analysis (Luinge & Veltink 2005). Accelerometer sensors are capable of measuring the movement in two or three dimensions as an inclinometer, demonstrating tilt of an object with respect to gravity (Goodvin et al. 2006). The development small devices continued rapidly, particularly the technology of micro-electromechanical systems (MEMS), has produced commercially as small inertial motion sensors suitable for human orientation and posture measurements (Goodvin et al. 2006). Inertial sensors have been used traditionally in aviation and robotics (Barshan and Durrant-Whyte 1995). Group of accelerometers and gyroscopes used to monitor the dimensional parameters; for instance, linear accelerations, rotational velocities in roll, pitch and yaw axes (Goodvin et al. 2006). Inertial sensors composed of MEMS accelerometers and gyroscopes have been suggested to analyse dynamic movement of the human being, such as daily living activities (Foerster et al. 1999). Lower extremities have been assessed by inertial systems, particularly the knee joint (Williamson and Andrews 2001). Inertial sensors have succeeded in measuring the movement of the lumbar spine (Hummel et al. 2005); however, inertial sensors using gyroscopes are usually subjected to a problem which may influence accuracy of orientation measurement and angular velocity. This integration system carries drift error which causes constraints in its usage, due to inaccurate measurements in less than a minute (Curtis et al. 1993). To correct the drifting errors, there are solutions which have been developed, such as the integration of magnetometers into the inertial measurement system in order to produce an absolute reference of magnetic north to reduce drift.

Recently, the gravitational acceleration become the dominant technique used for human motions, and can be used as a tilt (gravitational) sensors.

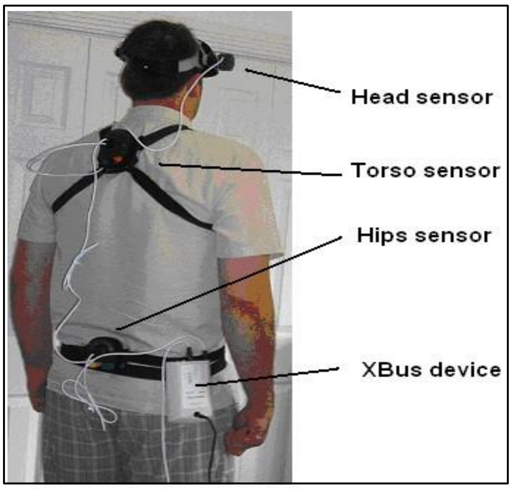


Figure 2.4.5: Spinal measurement system (Xsens MT9 inertial measurement) set-up for subject testing (Goodvin et al. 2006).

Currently, inertial sensors (accelerometers, magnetometers and gyroscopes) have the potential to be used to assess human movement. Authors have reported that these measures rely on accelerometers (Moe-Nilssen & Helbostad 2005; Jämsä et al. 2006; Kavanagh et al. 2006) or gyroscopes (Coley et al. 2005; Lee et al. 2003) or both (Saber-Sheikh et al. 2010). These sets of sensors have also recorded good accuracy with human motion studies (Boonstra et al. 2006; Jasiewicz et al. 2006). The inertial sensors are a viable method for quantifying cervical (Theobald et al. 2012), lumbar spine (Ha et al. 2013), knee (Brennan et al. 2011) and hip (Saber-Sheikh et al. 2010) range of motion.

However, gyroscopes are affected by drift in signals over time, whereas accelerometers and magnetometers can in theory be used to effectively correct this gyroscopic drift (Luinge & Veltink 2005).

Tri-axial accelerometers' level of accuracy has been stated at 1.3º for angular error with a reproducibility of 0.2º (Hansson et al., 2001). It was also compared with rotation alignment system and the correlation coefficient was found to be more than 0.99, which suggests that they are highly reliable (Wong & Wong, 2008). This system has been used successfully to observe the movement in terms of specific activities of everyday life and can be determined only from the training data (Kang et al., 2010). Additionally, such a system has been used to measure spinal motion and posture (Aloglah et al., 2010). One kind of accelerometers is piezoelectric accelerometer (sensing element) which creates a signal when some force is applied to it. This force is yielded by the inertia of the mass on top of the crystal as it is accelerated by some motion which is to be measured. Even though their name, accelerometers do not measure acceleration directly. They quantify the force applied, through the sensing element, and acceleration can be calculated through Newton's Second Law of Motion (F = m x a). It should be noted that this is only one type of accelerometer and there are other types that use capacitors as well as simple mechanical accelerometers but the way in which acceleration is calculated is the same; through Newton's Second Law of Motion.

A new version of the gravitational acceleration sensors (i.e. 3A sensors) has been developed for capturing the human motions relative to gravity. The system is sufficiently small and cost-effective for multiple sensors to be used along the length of the spine. Tri-axil accelerometer sensors are able to track absolute motion of spine over time, but the feasibility of using it in real-time for full spinal motion has not been explored by other researchers.

2.5 Summary

2.5.1 Contribution of multi-spinal regions

The value of understanding the motion between anatomical regions of the spine is becoming increasingly important. Authors have, however, commented that it is not sufficient to consider the thoracic spine as a single rigid body when evaluating overall spinal range of motion. It is thus becoming necessary to focus on a greater number of regions, rather than measuring only the three main areas of the spine (i.e. cervical, thoracic and lumbar). Furthermore, separate spinal regions are often studied in isolation;

however, clinically it is often the aim to assess more than one anatomical region. Therefore, there remains a need for a clinical device to simultaneously capture the kinematics of multiple small regions of the spine as well as hip movement behaviours.

2.5.2 Contribution of the upper and lower lumbar spine, relative to hip motion, in dominant daily sagittal tasks

The lumbar-hip complex range of motion is not only a criterion for regional impairment; another is the velocity of motion when performing these tasks, which can provide important information regarding movement behaviour and the inter-relationship between hip movement and lumbar. Behaviour movement of the spine-hips complex during sagittal functional tasks, such as flexion/extension, is a routine clinical protocol used to observe lumbar impairments (Porter and Wilkinson, 1997; McClure et al., 1997; Esola et al., 1996). Previously, clinicians have measured the lumbar spinal range of motion as one region, particularly flexion and extension in the sagittal plane. The traditional single 'joint' regions would fail to identify such region motions and may, therefore, oversimplify the description of movement (Lenzi et al., 2003). Therefore, it remains unclear whether additional, useful information can be gained by using a multi-spinal region instead of a traditional single joint (i.e. whole lumbar). A number of these studies have advised that investigating the lumbar spine as two separate regions may produce different movement behaviour compared with a traditional single joint region. In recent years, studies have focused more on the kinematics of the lumbar spine in more than one region for clinical subjects (Williams et al., 2013) and healthy subjects (Parkinson et al., 2013), identifying differences in regional contribution. No study has yet, however, considered multi-spinal regions of lumbar (i.e. upper and lower lumbar spine) versus hip movement, and across a series of dominant daily tasks.

Investigating daily dominant tasks with a novel technique of using a multi-spinal region system could provide new information for clinicians in order to assess the behaviour of lumbar spine movement during different tasks. Furthermore, the relationship between separate regions of lumbar and hip movement behaviour is still unknown for daily functional activities such as flexion, extension, lifting, sit-to-stand and stand-to-sit, particularly among the healthy population. A clinical evaluation of the lumbar-hip complex offers routine tests in chiropractic, osteopathic and physiotherapy clinics (Brantingham et al., 2012; Dankaerts et al., 2006). The evaluation of the behaviour of the spine and hip during spinal motions such as flexion/extension is one potential test advised to observe lumbar impairments (McClure et al., 1997; Porter and Wilkinson, 1997; Marcia A Esola et al., 1996). Clinicians utilise the results of motion tests, such as forward flexion, to aid in the clinical reasoning process when attempting to determine treatment and rehabilitation options.

Disorders of the lumbar-hip complex have been shown to affect lumbar spine and hip range of motion, as well as the interaction between these two anatomical regions (Murphy et al., 2006; Pearcy et al., 1985; Mellin, 1990; Esola et al., 1996). Moreover, disorders of the lumbar-hip complex in cardinal range of motion (lumbar flexion/extension) have a demonstrably significant effect on movement velocity, both at the hip and the lumbar spine (Shum et al., 2007b; Novy et al., 1999; Marras and Wongsam, 1986; Williams et al., 2013; Shum et al., 2007a; Shum et al., 2005a). This suggests that disorders of the lumbar-hip complex may affect functional tasks as well as the cardinal movements often employed in clinics.

2.5.3 The correlation of lumbar-hip kinematics between flexion and other functional tasks

It is not currently well-understood to what degree the cardinal motions, such as forward flexion, are related to more functional tasks. It is entirely possible that there is no relationship between forward flexion and other sagittally-dominant functional tasks, such as lifting, stand-to-sit or sit-to-stand. If this were the case, using forward flexion as a basis for exploring sagittal movement behaviour would be flawed, potentially leading to erroneous clinical judgements and reasoning. It may be the case, however, that forward flexion is closely related to other sagittal tasks, making the assessment of many tasks within the clinic unnecessary. Therefore, a better understanding of this relationship will aid in the interpretation of clinical assessment and treatment decision making.

Exploring the relationship between the kinematic profiles of flexion and three sagittally-dominant functional tasks (lifting, stand-to-sit and sit-to-stand) may yield new information. It may be the kinematic profile for the anatomical regions of upper and lower lumbar spine and hip, which is used in determining correlations between forward flexion and others closely related to other sagittal tasks.

Further conclusions of previous studies, which measured the spinal movement conducted to obtain the relative contribution of multi-spinal regions in dominant tasks during daily activities need an appropriate measurement system. This system should have the capability to track the absolute motion of the spine over time. It has bene suggested that tri-axial accelerometer sensors are useful for such measurements. This system can measure orientation, velocity and acceleration making such a system a viable option for a clinical assessment of a multi-regional range of motion. However, such a system needs to be evaluated against a range of spinal measurement systems. If this system is superior to other systems based on specific criteria, it is also compulsory to confirm its validity against a gold standard system and demonstrate its reliability in spinal movement.

Chapter 3: Methods

3 Methods

3.1 Study Protocols

This chapter describes the protocols and examination methods undertaken in this thesis.

3.1.1 Selection process of spinal motion analysis system

This section describes the criteria for the selection process of the spinal motion analysis system. In this section, various spinal measurement systems are evaluated including the tri-axial accelerometer sensors, which had been put forward as an appropriate system capable of measuring the spinal kinematics over time.

3.1.2 Programming methods

Based on evaluation process for measurement systems and having selected the tri-axial accelerometer sensors for measuring the spinal kinematics in experimental studies, this section will describe this nature of the system, axes orientations, sizes and programming methods (i.e. installation process, references axes of tri-axial accelerometer sensors and series of calibration methods).

3.1.3 Methods of examination the validity of tri-axial accelerometer sensors

This section will describe methodological examination of tri-axial accelerometer sensors against the gold standard system, the Rolly table. This section contains an introduction which summarise the rationale to this study and outlines the procedure used to evaluate the correlation and explains how data has been processed.

3.1.4 Methods of investigating the reliability of 3A sensors in quantifying multi-regional spinal range of motion

This section will describe the reliability examination of tri-axial accelerometer sensors by measuring the range of motion of five adjacent regions spanning the entire thoracolumbar and head-cervical regions. Two procedures were used in this section: procedure one was applied to measure the head-cervical region; and procedure two was used for five regions of thoracolumbar spine as well as participant's inclusion and exclusion criteria and the data analysis. This section will also explain data processing which is used to achieve the second aim of this thesis. The data which was obtained for reliability examination will help to analyse a novel technique that measures the relative contribution of regions from within the thoracolumbar and head-cervical during flexion, extension, lateral flexion to right and to left, and rotation to right and to left.

4.1.5 Experimental methods of lumbar spine and hip biomechanics during dominant daily tasks

This section will describe the methodological examination of study that measured the relative movement of the upper, lower, whole and lumbar spine regions with relative to hip during dominant daily functional tasks. This section will explain the processes of lumbar spine dividing, participant's inclusion and exclusion criteria and the data analysis with different statistic ways to explore motion and velocity magnitude using a traditional region of the lumbar spine as one single joint and to compare this with a sectioned approach. This section also describes data processing used for measuring the correlation between the kinematic profiles of flexion and three sagittally dominant functional tasks (lifting, stand-to-sit and sit-to-stand).

3.1.1 Selection process of spinal motion analysis system

The vertebrae are influenced by a complex physiological framework consisting of muscles, tendons, and ligaments and it is a complicated task to capture human spinal motion. The accurate 3D motion capture of the human spine is a complex procedure, particularly when real-time data is required. Assessing the range of motion and velocity of the hip, multiple spinal regions and the regional relationship during daily functional activities needs an appropriate system that has the ability to measure spinal kinematics over time.

Numerous invasive and non-invasive measurement methods have been used to measure spinal kinematics. Spinal kinematics have been measured by opto-electronic systems, radiology systems, electromagnetic systems, goniometers and inclinometers and inertial sensors. Opto-electronic methods have been used to measure range of motion in three dimensions for the cervical (Edmondston et al. 2007) and the thoracic (Edmondston et al. 2007); however, time-consuming and data processing can be complex (Ha et al. 2013). In research, an opto-eletronic system is able to track markers in space and can be to capture dynamic spine motion. It is, however, costly, and calibration, capture environment and the difficulty of moving such a system limit it to the laboratory setting and not ideally suited to the needs of clinical motion analysis.

Radiology involves filming the subject for the initial and final positions and is therefore, again, a static measurement technique as well as the radiation hazards constrain its applications.

Electromagnetic tracking devices are used widely in research as they are able to measure relative joint motion but are highly prone to electromagnetic interference and it is also a laboratory system.

In physical therapy and orthopaedic clinics, clinical examination of the spine often involves observing movements or using techniques that provide limited information, such as the inclinometer and finger-tip-to-floor tape measure. Simple clinical methods are only able to provide data from a single point in time and unable to measure movement behaviour across time (Williams et al. 2013).

In the midst of all these different systems that measure human movement, it is becoming increasingly important to choose an appropriate system which can overcome previous limitations and is appropriate for its specific application. Finding a system, however, which is portable, sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics and has the ability to measure movement in real-time is not a simple process.

One possible solution to measure the spinal movement over time would be to use accelerometers. The sensors are sufficiently small and cost-effective for multiple sensors to be used along the length of the spine. Tri-axial accelerometers measure linear acceleration along 3-orthogonal axes; however, they also measure tilt relative to gravity following the pendulum principle. This tilt measurement can be used to measure the orientation of the spine (Chang et al. 2009). This is only possible at low accelerations (Luinge & Veltink 2004; Luinge & Veltink 2005).

3.1.1.1 Device Selection Criteria

There are a number of systems capable of measuring spinal kinematics. There are specific aspects of the measurement methods that are important and, therefore, criteria of selection were developed in order to determine the most appropriate methods for the given application. The criteria included:

1. Portability (Y=20; N=5)
2. Number of dimensions measured (1D=5; 2D=10; 3D=20)
3. Is it possible to measure curvature? (Y=20; N=5)
4. Can multiple segments be investigated? (Y=20; N=5)
5. Cost (Exorbitant=5; Expensive=10; Inexpensive=20)
6. Is a line of sight required? (Y=5; N=20)
7. Dynamic measurements? (Y=20; N=5)

Each criteria was given a score out of 20 depending on each system's capabilities and how important it was for the given application. If a particular feature could not be scored in a linear scale and could only be given a ‘yes’ or ‘no’ option, the score was 20 if the technique had the capability and 5 if it did not. For the dimension's column, a score of 5 was given to a system that could only measure one dimension at any given time, a score of 10 to those that could measure two dimensions simultaneously and a score of 20 to those that could measure all three simultaneously. For the cost’s column, a score of 5 was given to a systems those are exorbitant, a score of 10 to those are expensive and a score of 20 to those are inexpensive. The scores were then calculated for each device in order to determine which device was best for the desired application.

Table 3.2.1: Selection matrix of main techniques used for spinal measurements.

Motion Analysis Technique	**Portability (Y=20; N=5)**	**Dimensions (1D=5; 2D=10; 3D=20)**	**Curvature (Y=20; N=5)**	**Analysis of Multiple Segments (Y=20; N=5)**	**Cost (Exorbitant=5; Expensive=10; Inexpensive=20)**	**Line of Sight Required? (Y=5; N=20)**	**Dynamic measurements? (Y=20; N=5)**	**Total**
Inclinometers	20	5	5	5	20	20	5	80
Electro-gonimeter	20	10	5	5	20	20	20	100
Electromagnetic	5	20	20	20	10	20	20	115
Optical tracking systems	5	20	20	20	5	5	20	95
Spine Mouse	20	10	20	20	10	20	5	105
Tape Method	20	5	5	5	20	20	5	80
3DMA (Zebris-US)	5	20	20	20	10	5	20	100
MRI	5	20	5	20	5	5	5	65
Radiography	5	5	20	20	5	5	5	65
Photography	20	5	20	20	20	5	5	95
Tri-accelerometer (THETAmetrix 3A Sensor Arrays system)	20	10	20	20	10	20	20	120

From table 3.2.I, the tri-axial accelerometer sensor system recorded the highest value (120) which was superior to the most common systems used for measuring spine movement. The tri-axial accelerometer is small enough and sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics. Such a new system needs to be identified, calibrated, and validated against a known output of orientation in order to confirm its reliability for measuring orientation.

3.1.2 Programming methods

3.1.2.1 What is tri-axial accelerometers sensors (3A)?

This system is a THETAmetrix 3A Sensor Arrays system is a highly accurate system which develop to calculate the orientations acceleration, roll, pitch and head axis. The portable 3A system comprises front-end software running on a PC and a main processor unit (MPU) which is connected to the PC by a ‘mini’ Universal Serial Bus (USB) cable). The MPU collates the data and transmits it to the Pearl Sensor software (included) running on a Windows PC or laptop. The sensor network strings between six 3A sensors connected along a single cable, which is permanently attached to the MPU (Figure 3.3.1). To make the system ready for use, the MPU must be plugged into the PC using a USB cable. The software will be automatically detected and connected to the MPU once it is plugged into the PC. The 3A system uses a three axis accelerometer sensors to measure the inclination in two dimensions. This system is a string of sensors that are wired in a ‘daisy-chain’ configuration with each sensor’s footprint measuring 24 mm2 (Theta-Metrix, Waterlooville, UK). Accelerometer contains a proof mass element mounted on a fixed base with strain sensitive wires attached. Increased acceleration results in increased deformation of the mass element causing change in the strain in the wires. They measure linear acceleration along the sensing axes based on the equation, force = mass x acceleration. Orientation when static may be measured by functioning as inclinometers, measuring tilting angle with respect to gravity and it is this function that is commonly used in spinal motion analysis (Williams 2011). Sensors are using to measure angulation movement and velocity over time because accelerometers are providing axial acceleration data pertaining to absolute orientation (tilt), with respect to gravity.

Having collected the data of absolute orientation, described as Euler angles, this was converted into rotation matrices and the resultant angles between two adjacent sensors were calculated through matrix multiplication to determine the motion of each individual spinal segment (Swaminathan et al. 2016), through a custom written code in Matlab (Williams et al. 2013).

The 3A system has advantages and disadvantages, which will be outlined below.

Advantages:

I. Highly accurate for angular measurements (roughly 1 degree) compared to other systems such as opto-electronic systems. The level of accuracy for accelerometers has been stated at 1.3º for angular error with a reproducibility of 0.2º (Hansson et al., 2001). It was also compared with rotation alignment system where the correlation coefficient was found to be more than 0.99, which suggests strong reliability (Wong & Wong, 2008).

II. They have previously been used in biomechanics for posture analysis and the assessment of neck pain. This system has been used successfully to observe movement in terms of specific activities of everyday life and can be determined only from the training data (Kang et al., 2010). Moreover, similar system principles has been used to measure spinal motion and posture (Aloglah et al., 2010).

I. Can measure orientation in two dimensions simultaneously and biaxial accelerometers were shown to be highly reliable when compared with goniometric and electromagnetic systems (Wong et al., 2009).

II. 3A system is sufficiently small and cost-effective for multiple sensors to be used along the length of the spine and do not suffer the drift problems seen with gyroscopes, making them usable over prolonged periods such as for postural analysis (Breen et al., 2009).

Disadvantages:

I. A disadvantage of using multiple sensors in a string is that the complexity of analysis would increase greatly.

II. Clinically, it cannot measure the spinal kinematics during the rotation movement; however, this can be addressed by measuring the subject from a side-lying position when rotating the spine to the right and then to the left (Alqhtani et al., 2015).

Figure 3.3.1: A portable set of six sensors, linked in a 'daisy chain' formation, which comprise 3A sensors and measure orientation and acceleration relative to gravity.

3.1.2.2 Installation process of 3A sensors

The tri-axial accelerometer sensors system comprises front-end software running on a PC. The control software should be unzipped into a convenient place where it can be easily found e.g. "C:\Program Files\Pearl Sensors". A shortcut placed on the desktop or in the menu is also advised for convenience. Once the software is installed, it must be configured before the system can be used. When selecting the number of sensors (Figure 3.3.2), the sensors will need to have their addresses assigned while each sensor in the array needs a unique address.

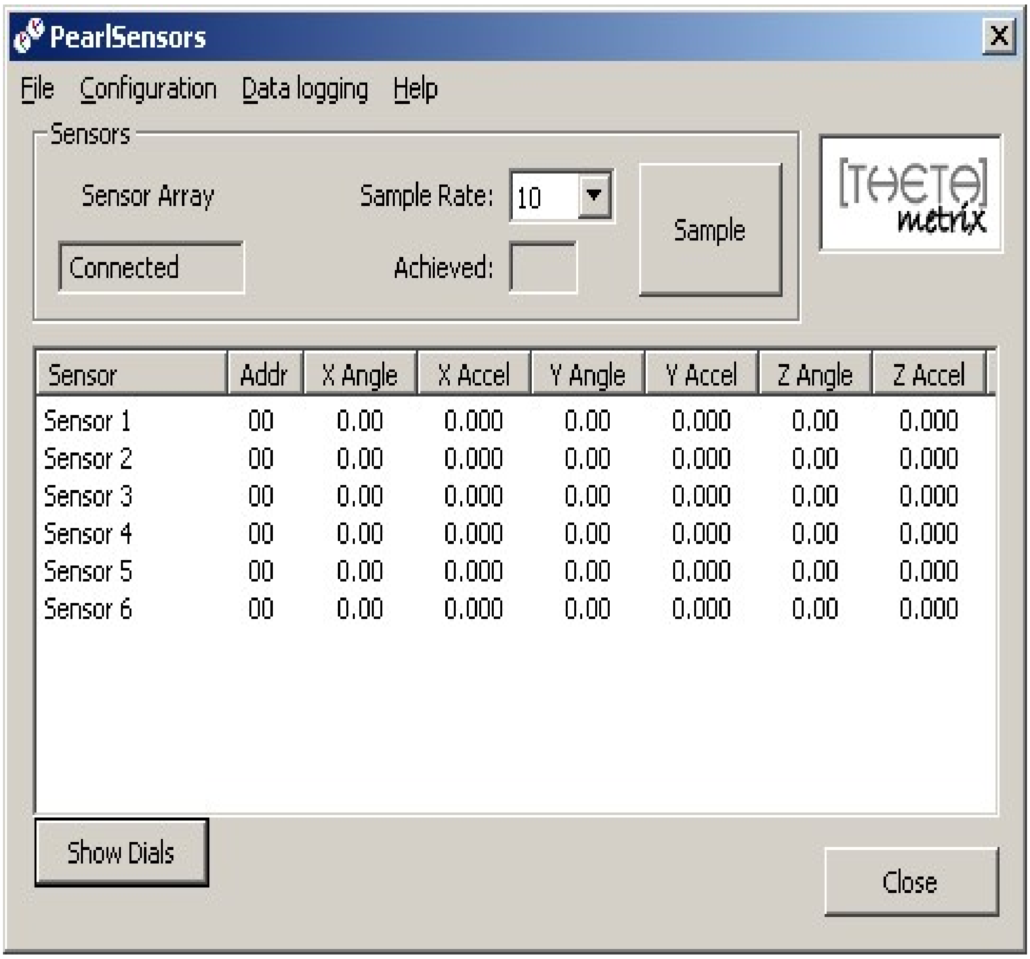


Figure 3.3.2: The main window of 3A sensors.

3.1.2.3 Reference axis of Tri-axial accelerometer sensors

The select reference axis allows the user to choose which of the three sensor axes is used as the vertical axis for measuring angles and acceleration (Figure 3.3.3). When the sensor is placed on a horizontal surface with the label facing upwards and the cable pointing towards the user's left, axis A is horizontal and perpendicular to the cable (positive towards the examiner), axis B is horizontal and aligned with the cable (positive to the examiner's right) while axis C is vertical up/down (positive up).

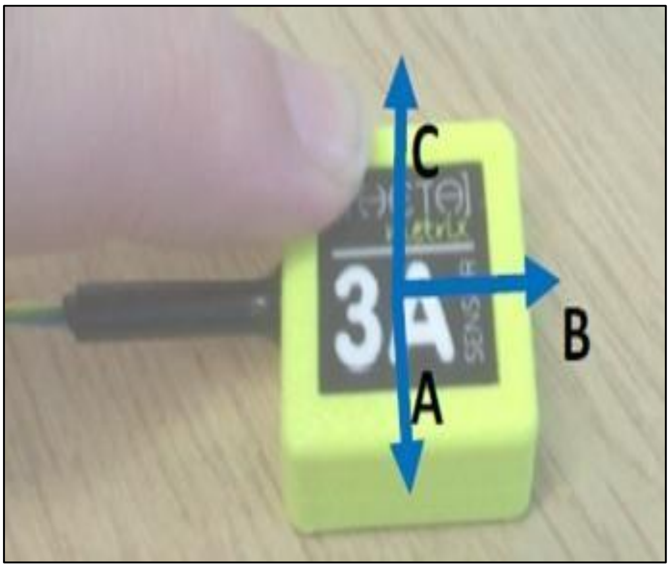


Figure 3.3.3: A, B and C reference axes

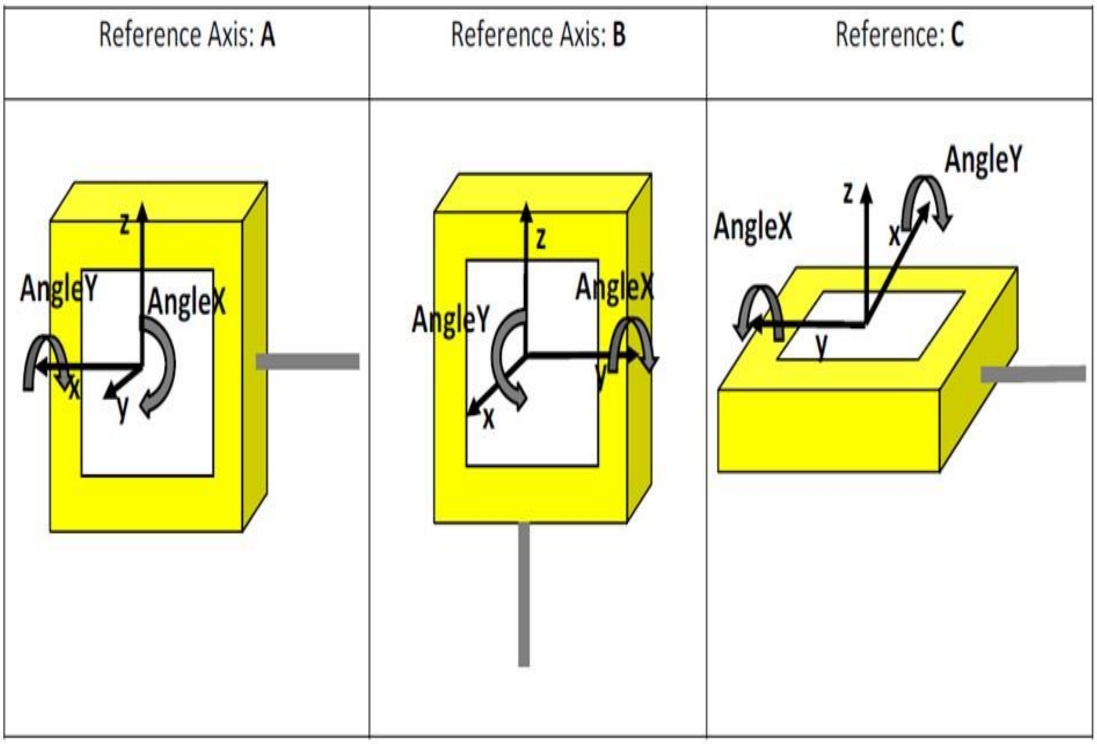


Figure 3.3.4: Illustrates the definition of the x, y and z axes when the reference axis is set to A, B and C. The AngleX and AngleY rotations are the roll and pitch angles, while the

AngleZ value is the angle between the reference axis and the vertical downward axis (i.e. gravity).

Axis C is possibly the most suitable when the sensors are horizontal and, as shown in the guide (Theta-Metrix, Waterlooville, UK) axis A or B is recommended if the sensors are to be used in the vertical plane such as for human spinal measurements. Depending on the easiest and most suitable location for the cable, B has been selected for the experimental measurements of this research.

3.1.2.4 Calibrating a sensor of Tri-axial accelerometer sensors

Each sensor must be calibrated individually, displaying its name (i.e. sensor 1, sensor 2 etc.) and the per axis calibration for data offset and gain. It is important to calibrate these acceleration values correctly, as they are used to calculate the orientation angles (angles off vertical) for each axis. The first step in calibrating a sensor is to zero the current offsets and gains. The offset value for each axis is obtained by measuring twice for increased accuracy; each measurement is taken with the sensor in a different orientations, which will average out any slope in the table or bench surface. The accelerometers used in the tri-axial accelerometer sensors units should give a raw value of 1024 to represent 1g; however they do not quite do this, so it is necessary to map this inconsistency to obtain more accurate measurements from the sensor. This error in range is called the gain. Therefore, two procedures of calibration will apply: offset and gain calibration.

The offset calibration process must reset the offset button and then reset the gains button, which will zero out any current calibration and set the offsets to zero.

Steps of measuring the offset values for each axis are as follows:

1) The researcher holds the sensor against a flat surface as shown in figure (3.3.6) to calibrate axis A offsets (sample A), by pressing the start button for sample A, holding the

sensor steady until the value for standard deviation (SD) drops ideally to below 0.5 before pressing stop. Then the same sensor was rotated 180° to calibrate (sample B) against a flat surface in the orientations shown in figure 3.3.6.

2) Rotating the sensor by 90° to the left and then to the right in order to calibrate axis B offsets (samples A and B), and the researcher pressed the start button for sample A and then sample B by holding the sensor steady until the value for standard deviation (SD) drops ideally to below 0.5 before pressing stop.

3) The sensor placed on vertical to calibrate C axis as shown in figure 3.3.6, and the researcher pressed the start button for sample A and then sample B by holding the sensor steady until the value for standard deviation (SD) drops ideally to below 0.5 before pressing stop.

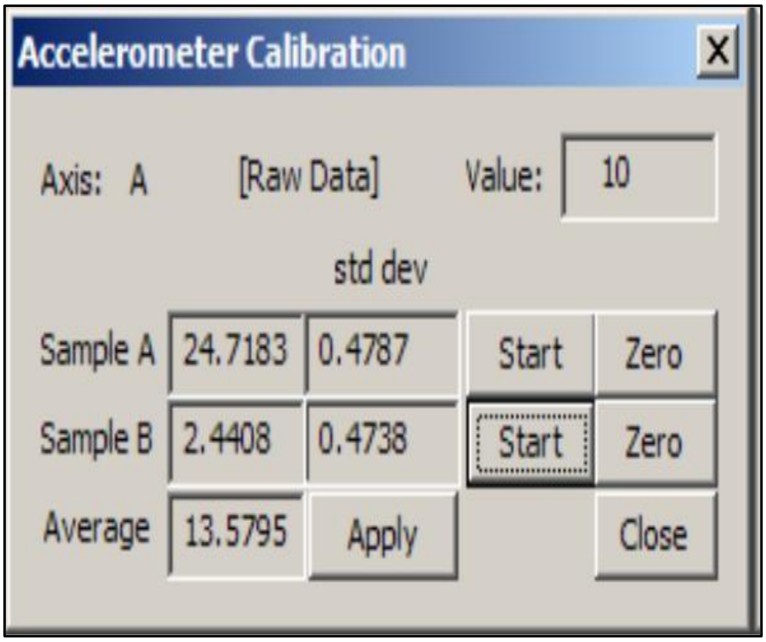
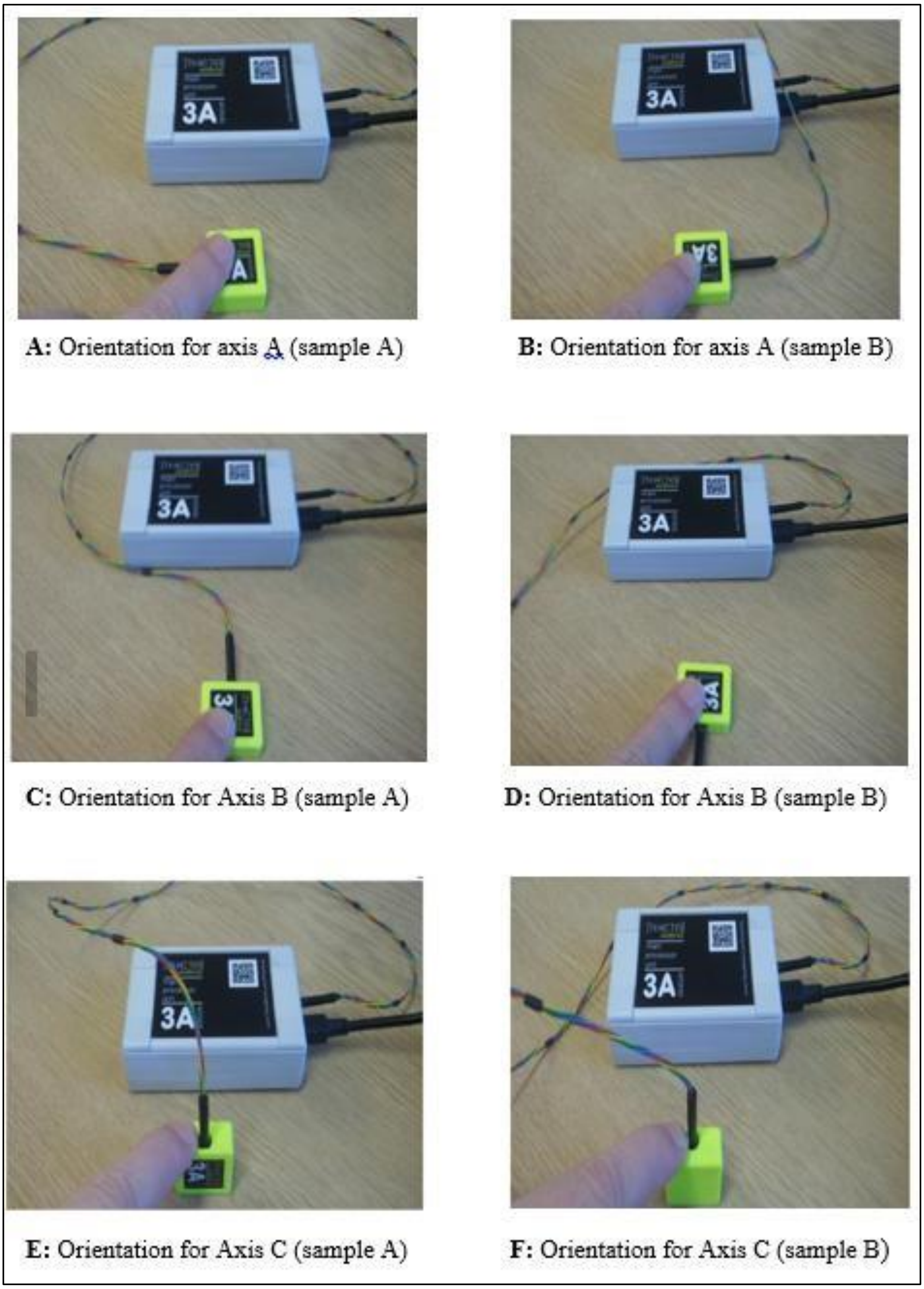


Figure 3.3.5: Orientation for axes.



A: Orientation for axis A (sample A)

B: Orientation for axis A (sample B)

C: Orientation for Axis B (sample A)

D: Orientation for Axis B (sample B)

E: Orientation for Axis C (sample A)

F: Orientation for Axis C (sample B)

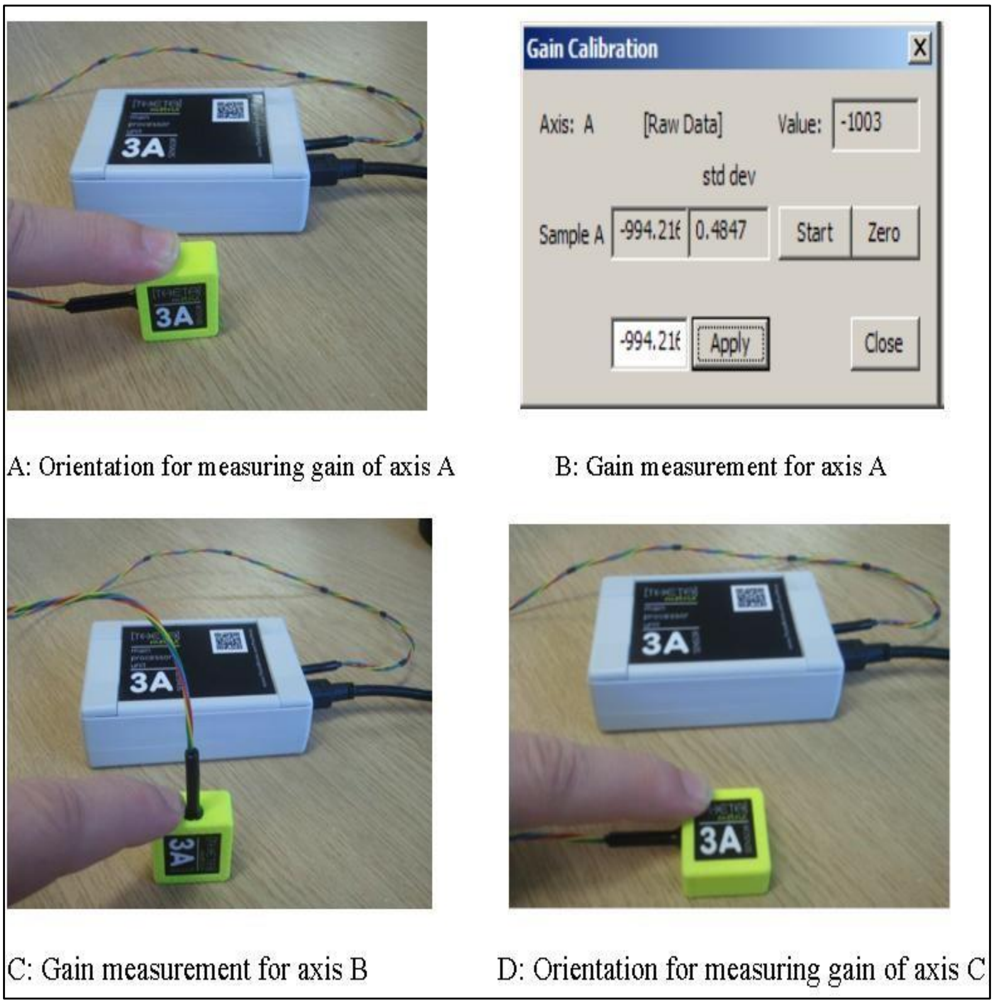
Figure 3.3.6: Positions of sensors calibration.

3.1.2.5 Calibration of gains

Calibration of the gain is very similar to the offset measurements, except that only one set of measurements is taken per axis, which calibrate as follows:

Holding the sensor so that the axis is measured and is aligned with gravity (i.e. vertical), press the start button to begin the measurement until SD falls below 0.5 in order to measure the gain of axis A (figure 3.3.5).

Repeat the measurement process for axes B and C with the sensor in the orientation as shown in figure (3.3.7).



A: Orientation for measuring gain of axis A

B: Gain measurement for axis A

C: Gain measurement for axis B

D: Orientation for measuring gain of axis C

Figure 3.3.7: Sensors calibration using gain positions.

For the offset calibration, the gain values are a production artefact and should not be subject to drift. The gain values are recorded in the software configuration file and the sensor unit will be fully calibrated. The process of this calibration was repeated for all sensors in the sensor array.

3.1.2.6 Displaying Data

At this stage, the system is ready to be used for data gathering with the main display showing a configured and calibrated system. There are several options available from the main window. The Dials button will show or hide a dial display for the currently selected sensor. There are four traces per scope display and each trace may show any of the available data elements within the system. Each trace may have its sensitivity (range) and centre altered individually.

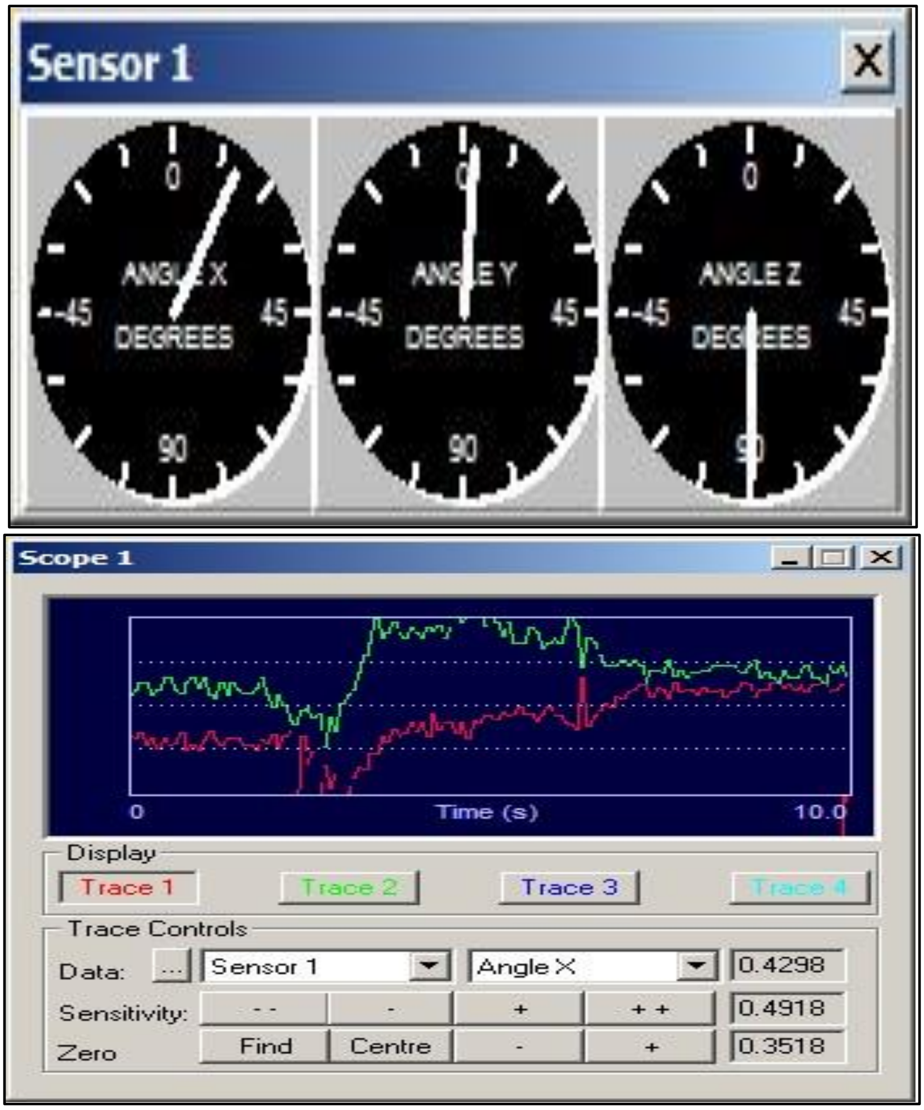


Figure 3.3.8: The dial display for sensor 1 and a "Scope" display showing two data traces.

At this stage, the nature of the tri-axial accelerometer sensors device, selection process and calibration processes are known by the accelerometers motion shown on the dials or scope displays. The next stage is an evaluation of the validity of tri-axial accelerometer sensors against a gold standard device in terms of accuracy level when measuring pitch, roll axes and cross-talk measurement at different inclination degrees for pitch and roll axes.

3.1.3 Methods of examination the validity of 3A sensors

In the biomechanics field, inertial measurements of motion (Roetenberg 2006) is developing to overcome the limitation of optical or magnetic 3D motion tracking systems for the measurement of mobility impairments (Schulze et al. 2012; Cutti et al. 2010; Ferrari et al. 2010; Cutti et al. 2008). One of the most important reasons for using internal sensors in biomechanics is that it permits evaluation of the function motion in real-time and conditions with fewer limitations compared with optical or electro-magnetometer systems.

Tri-axial accelerometer sensors are capable of measuring movement in real-time; portability, non-invasive application, minimal system footprint, and sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics. This system that uses tri-accelerometers sensors has been selected based on specific criteria in table (3.2.1). It has recorded the highest value, which mean this system could overcome the most limitations of spinal measurement systems. However, it still needs to confirm its validity against a “gold standard system”.

3.1.3.1 Validity consideration

To examine the accuracy of the selected measurement system against a gold standard system, concurrent validity used to compare the level of accuracy between two systems. Such type of validity uses to compare the measurement data which obtained by new measurement system with measurement data which measured by a previously validated measure, often a gold standard measurement (Portney and Watkins, 2009).

Concurrent validity was used to evaluate 3A sensors against the gold standard system; a previously validated measure. This is deemed a well-known method of investigating the validity of a new instrument or device (Cortney and Watkins, 2009). Concurrent validity can be used when a new or unexamined instrument is easy and feasible to administer and is potentially more efficient while predictive validity refers to a tool's capability to predict and make inferences about the future.

3.1.3.1 Instrumentation

Whilst, 3A system is a new version of gravitational accelerometers family, it requires to validated against the gold standard system. Based on concurrent validity study conducted by Hole et al. (1995) to describe the method applied to obtain correlation of two instruments, inclinometers (gravity goniometers) and cervical range of motion system, a similar method will be used in the present study.

The gold standard which will use for this study is a highly precise system 'Rolly table' (Figure 3.4.1). The Rolly table has the capability for measuring the axes orientations that sensors move through. Rolly table which is composed of a 3-axes gimbal table that allows single or multi-axes trajectories of motions. The rotary table was custom made by ThetaMetrix. Its output was given by three digital encoders (model number: ERN-420) with one encoder used for each axis (roll, pitch heading). The encoders were manufactured by Heidenhahn (Sweden). Each encoder had 3,600 lines per revolution (360 degrees) with four steps per line giving steps of 1/40 of a degree. The lines were generated by fixed marks which were optically scanned by the encoders. The rotary table was manually rotated in each axis individually with different speeds being used (slow, medium, fast). This device is a gold standard system used for sensors' calibration and their validation at their place of manufacture at Theta-Metrix (Waterlooville, UK).

Figure 3.4.1: High precision rolly table (Jig)

3.1.3.3 Study Protocol

Six 3A sensors and a high precision rotary table were used to measure the orientation of each axis (roll, pitch, and heading). Roll is defined as rotation about the x-axis, pitch as rotation about the y-axis and heading as rotation about the z-axis. The rotary table determined orientation through the use of digital encoders (ERN-420, Heidenhahn, Sweden). Each encoder was accurate to 1/40 of a degree. Specifications of the digital encoders can be found in the manufacturer's guide (Heidenhain, 2013). The accuracy of the table was checked by rotating the table through 360° in each axis and checking the output of the digital encoders. This method was chosen to validate the 3A sensors as digital encoders are known to have a high accuracy for orientation output.

The 3A sensors were attached to the rotary table using double-sided adhesive tape. Double-sided adhesive tape is a widely-used method of attachment in kinematic analysis along with Coban tape. In this scenario, double-sided adhesive tape provided a much more secure attachment than Coban tape. Each sensor was attached to a piece of metal so that each had the same orientation so as to avoid problems with axis cross-talk during data analysis. Figure 3.4.2 shows the experimental setup of the sensors.

Once the sensors were secured, two axes of the rotary table were manually secured at 0° so no motion occurred for these two axes. Once fixed, data collection for the rotary table and 3A sensors started and the table was rotated through its full range of motion; that is, ±180°. The table had to be manually rotated and therefore the speed of rotation was difficult to keep constant. Having completed the trial for one axis, another two axes were fixed and the free axis was rotated through its full range. This was performed until all axes had been investigated. The process also included a number of trials to demonstrate cross-talk.

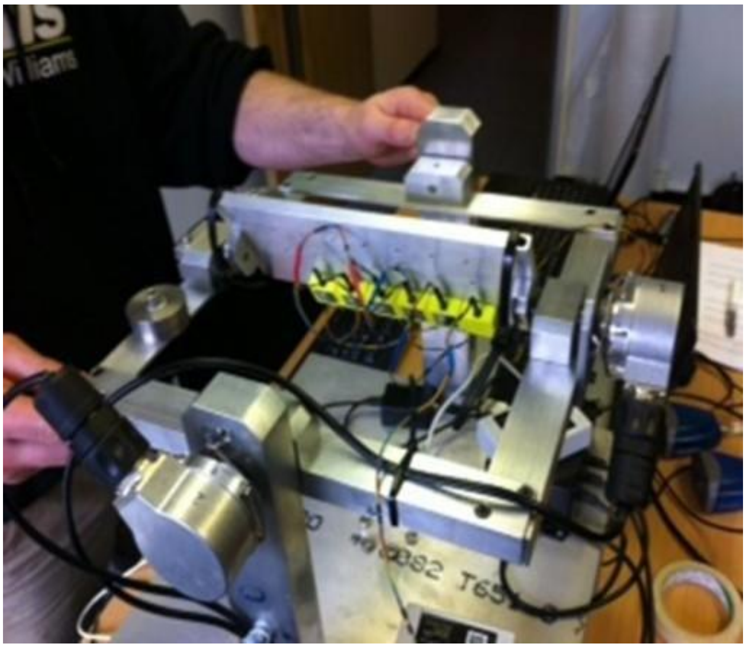


Figure 3.4.2: 3A sensors mounted on roll axis of jig

The protocol was written in two separate sections: first, the section tested the sensors to check their full range of motion in a single axis; second, the level of cross-talk in different axes was tested by tilting the sensors to a specific degree and then rotating them through a full range of motion, axis by axis.

The first part of the trials were set out as follows:

- The roll and heading axes were locked at 0˚.
- The rotary table was rotated in pitch through ±180˚.
- The pitch and heading axes were locked at 0˚ and the trials were carried out by rolling.
- This was repeated until each axis had been rotated through its full range.

The second part of the trial was set out as follows:

- The rotary table was positioned in 30˚ roll and this axis was locked.
- The rotary table was slowly rotated in pitch through ±180˚ being careful not to move in heading.
- The rotary table was positioned in 60˚ roll and this axis was locked.
- The rotary table was slowly rotated in pitch through ±180˚ being careful not to move in heading.
- The rotary table was positioned in 30˚ pitch and this axis was locked.
- The rotary table was slowly rotated in roll through ±180˚ being careful not to move in heading.
- The rotary table was positioned in 60˚ pitch and this axis was locked.
- The rotary table was slowly rotated in roll through ±180˚ being careful not to move in heading.

3.1.3.4 Data processing

All raw data obtained using both devices 3A system and RT system were processed using Excel 2010. The values of angles have changed from radian to degrees for two devices' axes (roll and pitch). Time has been normalised and changed to a percentage. Correlation between 3A sensors and RT axes were explored comparing two systems capturing data at each axis using Pearson correlation coefficient (r). The data was plotted on charts to reveal the correlation between the two devices. The magnitude of the root mean square error (RMSE) for each direction was obtained by calculating the difference between the angles' value of 3A and RT for each direction of movement (i.e. roll, pitch, roll, when pitch locked at 30 and 60° and pitch, when roll locked at 30 and 60°). The formulae used were RMSE= √sum (3A (i) -RT (i))^2)/N, where 3A (i) and RT (i) represent the 3A and RT angles and N, the number of comparison samples. The percentages of RMSE were obtained by using the formulae which were utilised by Brennan et al. (2011): Percentage (%) = RMSE/ total ROM (360°)*100, total of ±180˚. The correlation between two devices' measurements has been obtained using Excel 2010.

3.1.4 Methods of investigating the reliability of 3A sensors in quantifying multiregional spinal range of motion

The evidence from the validity study suggests that tri-axial accelerometer sensors are valid and capable of measuring spinal movement. However, the reliability of tri-axial accelerometer sensors has not yet been examined for spinal motion. Therefore, this study primarily aimed to evaluate the reliability of a novel, multi-accelerometer system, by measuring the range of motion of five adjacent regions spanning the entire spine as well as head-cervical movement. Secondly, this system was then used to consider the relative contribution of five regions from within the thoracolumbar region as well as the head-cervical contribution.

This section will explain the examining protocols on 18 healthy participants and the processing of the data. Methods of reliability examination in this section will compromise obtaining the relative contribution of five regions of thoracolumbar region as well as head-cervical region. Dividing the spinal region into five regions is a novel technique, which requires confirmation of the reliability of 3A sensors.

3.1.4.1 Reliability consideration

It is necessary to provide constant or reproducible values with tolerance errors of measurement when no variable is affecting the attribute that the measurement is quantifying (Rankin and Stokes 1998). Errors of measurement are normally one of two different types: one is systematic errors, which may constantly under-estimate or overestimate values. A systematic error does not hinder the reliability of the outcome being measured. However, systematic errors do generate problems of validity when the measured value is not the exact representation of the quantity measured. The second type of measurement error is random error. Random error does pose a problem of reliability, as it occurs due to unpredictable factors such as mechanical inaccuracy, lack of experience and fatigue. The unpredictability of the working field and subjects involved cannot be avoided, even if the errors' sources are expected. Random errors can be minimised when the performed numbers of measurements take their average value as a good estimate of the accurate value (Portney and Watkins 2009). This helps to ascertain the difference among the values measured by yielding a ratio called the reliability coefficient that has a coefficient of 1.0 for maximum reliability. Reliability coefficients are based on a measure of correlation such as Pearson's product moment correlation or the Intraclass Correlation Coefficient (ICC) and range between 0.00 and 1.00. The more reliable the measurement response, the less error variability there will be around the mean (Bruton et al. 2000). The ICC is the best approach that can be used to examine relative reliability between two or more trials. It is based on measures of variance from the analysis of variance (ANOVA) (Portney and Watkins 2009). There are different types of ICCs available such as equation ICC (3, 1), a two-way mixed model/absolute agreement, which was used to assess the reliability of a fixed rate for repeated measurements (Rankin and Stokes 1998). A coefficient below 0.4 is considered an indication of poor reliability, between 0.5 and 0.75 is considered as moderate to good reliability and above 0.75 suggest excellent reliability (Fleiss 1986). Reliability cannot rely solely on ICC, since it measures only the strength of association between the two variables but not the extent of agreement between them.

3.1.4.2 Instrumentation

Tri-axial accelerometer sensors were used to measure spinal range of motion. All sensors were connected to a laptop computer via a USB cable. The accuracy of the sensor string (‘3A sensors’) has previously been investigated within a high precision, controlled environment through the use of an ‘XYZ’ table (i.e. high precision yaw, pitch and roll movements). High correlation was reported when comparing the tri-axial accelerometer sensors and ‘table’ data ($r = 0.98$, root mean square errors = 0.70 - 1.39%) (Alqhtani et al. 2015). These measures describe the correlation and deviation of the 3A sensors, relative to the gold standard data.

3.1.4.3 Participants

A total of eighteen male participants were recruited (age = 30.6 ± 7.4 years; weight = 76.6 ± 7.4 kg; height = 171 ± 5.3 cm. This study was designed to explore further variables as references for physiotherapists to use during spinal assessment by comparing normal variables which were obtained in this study with pathological variables. This study faced difficulty in recruiting a combination of female and male subjects. All participants in this study were male. Meanwhile, a number of people agreed to participate and confirmed their participation by email but changed their mind and refused to attach the sensors to their skin at the experimental lab.

The cohort size was initially based upon a review of similar reliability studies (Williams et al. 2010) before its appropriateness was re-evaluated and confirmed, following statistical analysis of current study data.

Participants were in good health with no history of back pain or leg pain that may be attributed to the back within the last 12 months. participants were excluded if they had any history of spinal surgery, fracture, dislocation or any structural defects of vertebral structures, or any disorder affecting the cervical, thoracic or lumbar region. The study was approved by the Ethics Committee of Cardiff University and all participants provided written informed consent having been explicitly informed of the experimental procedures. Participants were recruited via a circular email to staff and postgraduate students, meaning our cohort was a convenience-based sample. All participants provided informed, written consent prior to their visit for data collection.

3.1.4.4 Procedures

When a participant attended the experimental place (Motion Lab) at the Cardiff School of Engineering, they were to return a signed consent form, which had been provided earlier. The participant should move to a private area to change their clothes and wear shorts. Each participant is advised to move their head-cervical and spine forward, backward, rotation and lateral flexion right and left from three to five times before placing the sensors on the body. These exercises have been used as a warm-up and to orientate subjects with the tasks involved. Spinal range of motion was assessed through the development of two protocols. Protocol one was devised to evaluate the reliability of the device for measuring cervical movements including flexion, extension, lateral flexion to right and to left, rotation to right and lift, before protocol two was implemented, which focused on using the device to investigate thoracolumbar range of motion.

3.1.4.4.1 Protocol one

After completely drying the skin, the participant is instructed to sit on a chair without back support and then place one sensor on the forehead and another on the skin overlying the spinous processes of first thoracic vertebra (T1). The two sensors define a region that quantifies head-cervical range of motion (Figure 3.5.1). Sensors were attached using double-sided tape and participants were asked to move their heads through full range of motion. The participants were instructed to look at a specific point (marker) on the wall in order to record measurements starting from a natural position. Flexion-extension and right-left lateral flexion were recorded with the individual while sitting down. As it was only possible to measure two planes of motion (due to inclination relative to gravity), axial rotation (right and left) was obtained from a prone position with the head protruding beyond the end of the treatment table (Alqhtani et al. 2015). Participants performed three repetitions of each movement.

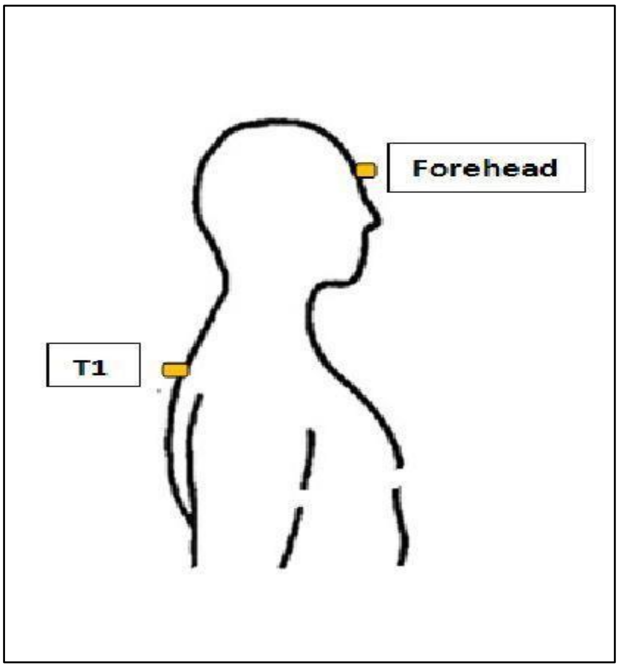


Figure 3.5.1: Schematic represents the location of forehead and T1 sensors.

3.1.4.4.2 Protocol two

After warming up, the researcher dried the back of the participant using a tissue in order to ensure excellent adhesion for sensors on the skin. The participant was then asked to lean forward on the table (waist level), in order to determine the specific spinous processes. For measurement of the thoracolumbar motion, six sensors were placed on the spinous processes of first thoracic vertebra (T1), fourth thoracic vertebra (T4), eighth thoracic vertebra (T8), twelfth thoracic vertebra (T12), the third lumbar vertebra (L3) and first sacral vertebra (S1). This method created five anatomical regions of interest: upper thoracic (UT); middle thoracic (MT), lower thoracic (LT), upper lumbar (UL) and lower lumbar (LL). The sensors were firmly attached to the skin of each participant for the relevant spinous processes using double-sided tape (Figure 3.5.2). T1 was located below the vertebra prominent (C7) and T4 by counting down three prominent vertebrae. This point was identified by detecting T3 which lies in the middle of a line drawn between the roots of the spine of each scapula (Clarkson 2000). The eighth thoracic vertebra was recognised by counting down four spinous processes and verified by locating T7 at the middle of a line drawn between the inferior angles of each scapula (Clarkson 2000). The twelve thoracic vertebra was located a further four spinous processes below. The researcher used his hands to hold the participant's waist, placing two index fingers on both iliac crests and two thumbs were extended to palpate the L4 spinous process, which was located at the same level of the iliac crests. The third lumbar vertebra was identified by counting upward one region. Two superior posterior iliac spines are usually easy to identify when the participant is asked to lean forward from a standing position; therefore, S1 was located at the same level of these two locations. Four sensors were attached to the skin's surface of spinous processes of the first, fourth, eighth and twelfth thoracic vertebra according to a previously established model and protocol (Willems et al. 1996) and two sensors were placed over the spinous processes of the third lumbar vertebra (L3) and the first sacral vertebra (S1) (Williams et al. 2010).

The participant was instructed to stand barefoot on assigned markers and focus on a wall marker set at a height of two metres, with arms relaxed by their side. The participant was asked to move their trunk into flexion-extension and right-left lateral flexion. As only two planes of motion were possible (due to inclination relative to gravity), axial rotation (right and left) was obtained from a side lying position, where the participant was asked to rotate their trunk to the right and left, while the researcher fixed their hip and lower extremities (Alqhtani et al. 2015). Starting position was standardized by using two reference kitties were fixed perpendicular to bed edge (90°), one at shoulders levels and another at pelvis level. To start the rotation, the skin of posterior aspect of shoulder (i.e. the prominent of scapula process) and the posterior aspect of pelvis (i.e. posterior-superior iliac spines area) should contact with two kitties. To obtain full rotational range of motion of thoracic-lumbar, the participant was instructed to rotate their head in the direction of movement with full horizontal abduction in shoulder. The participant performed three repetitions of each movement cycle.

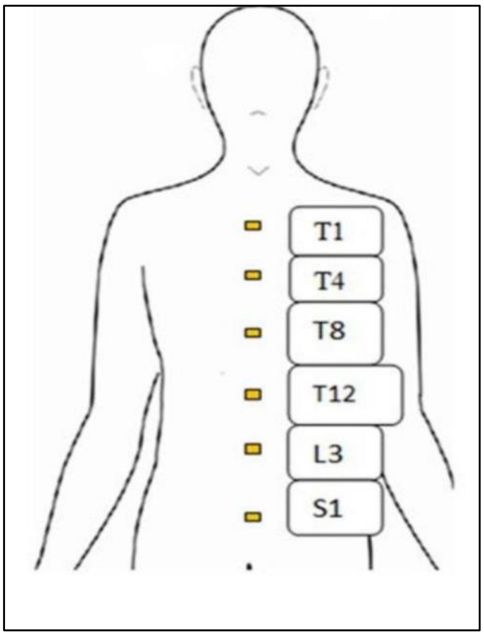


Figure 3.5.2: Schematic representation of the location of spinal sensors.

3.1.4.5 Data analysis

Data were collected at 30Hz and raw data were transferred to Matlab and filtered at 6Hz (low-pass, Butterworth) to remove high frequency noise (Scholz et al. 2001). Data were recorded at tilt angles relative to gravity (absolute angles) and regional ROM was defined as the relative motion between adjacent distal and proximal sensors (i.e. relative angles). Subsequently, regional spinal movement-time curves were generated for HC, UT, MT, LT, UL and LL from which peak range of motion values were calculated.

Matrix laboratory (Matlab-R2013a) was used to measure the relative motion during flexion-extension, right-left lateral flexion and axial rotation (right and left). Regional range of motion (i.e. HC, UT, MT, LT, UL and LL) = Peak ROM between upper and lower sensor, relative to each region.

Within-day, intra-tester mean values (one examiner; three tests) and reliability measures of multi-spinal regions ROM used during flexion, extension, and lateral flexion to right and to lift from upright standing and rotation to right and to lift from a side-lying position. Intra-class correlation coefficients (ICC) with 95% confidence intervals (Cl) were calculated using the Statistical Package for the Social Sciences (SPSS 20), to evaluate the repeatability of the three repetitions recorded for each movement. The ICC value is recognised to provide a measure of repeatability (Bruton et al. 2000) with values classified using the following thresholds: < 0.4 = poor, 0.4 - 0.75 = fair to good, >75 = excellent, as per Fleiss (1986).

The error measurement was used to define the extent of error, meaning that greater reliability is defined by a smaller error value (Bruton et al. 2000) because ICC values alone cannot be explained clinically as they do not provide any indication of the level of discrepancy between measurements (Rankin and Stokes, 1998). Therefore, standard error of measurement (SEM) and minimal detectable change (MDC) were calculated. Standard error of measurements was obtained using the formula: SEM = SD*(√1-ICC) (Denegar and Ball, 1993). Minimal detectable change was calculated using the formula: MDC= 1.96*(√2)*SEM (Kropmans et al., 1999).

3.1.5 Experimental methods of lumbar spine and hip biomechanics during dominant daily tasks

Indeed, the lumbar spine is a complex structure influenced by stress, compression, decompression, strain, and tension, which is caused by extensive force or malposition. However as there is a strong relationship between lumbar spine and hip movement, the basic anatomy of hip joint was explained. Lumbar-hip complex allows complex movements, which may contribute to pain that affects the lumbar region and adjacent joints' kinematics.

Sagittal tasks in daily living have been studied widely, however, researchers have not tried to examine multi-spinal regions with or without hip movement. They have examined the kinematics relationship between lumbar spine (as a single region) and hip motion on sagittal plane and have found diversity in the magnitude of movements. Even though this procedure (i.e. relationship between hip and lumbar as a single region) has been widely used, there is still a need to understand the kinematics of multi-spinal regions and the relationship with the hip during common daily functional activities. In recent years, a number of studies have called for further investigation into the function of the lumbar spine as two regions (upper lumbar and lower lumbar spine) when performing different functional tasks. However, the concept of considering the motion and function of the lumbar spine in terms of lower lumbar and upper lumbar regions has not been widely investigated either in clinical studies or with healthy subjects. More recently, some studies have measured the lumbar spine as two separate regions; however, these studies were limited by measuring the static positions regardless of dynamic motion over time.

Unfortunately, it is not well known to what degree the fundamental movements, such as spinal flexion, are related to more daily functional tasks. It is completely possible that there is no relationship between spinal flexion and other sagittal functional tasks, such as lifting, stand-to-sit or sit-to-stand. If this is so, operating spinal forward flexion as a basis for known sagittal movement behaviour would be questionable, potentially leading to flawed clinical judgements.

The lumbar spine has divided into two regions (whole lumbar and lower lumbar spine) in order to measure the lumbar curvature in lower back pain patients (Williams et al. 2010; Williams et al. 2012). However, Learding et al. (2011) and Mitchell et al. (2008) have used different regions when they divided the lumbar spine into two regions (upper and lower lumbar spine). Williams et al. (2010) and Williams et al. (2012) have conducted their studies on the clinical population, while, Learding et al. (2011) and Mitchell et al. (2008) have conducted their studies in healthy subjects. Learding et al. (2011) and Mitchell et al. (2008) have been limited to measuring only the start and end range of motion points (i.e. the dynamic movement over time has not been obtained). Furthermore, no study has yet examined the range of motion of upper lumbar spine and lower lumbar spine relative to hip movement, which associated with lumbar problems.

The aim of this study was to determine whether dividing the lumbar spine as two separate regions yields a different understanding of the movement behaviour of the spine, compared to a traditional single joint region. This was determined by exploring motion, using a traditional region of the lumbar spine as one single joint (S1 to T12) and comparing this with a sectioned approach, where the lumbar spine was divided as two distinct regions, namely the upper (L3-T12) and lower (S1- L3). Three regions of lumbar spine will be normalised to region s (i.e. upper lumbar/3 vertebrae, lower lumbar/3 vertebrae and whole lumbar/6 vertebrae). The novel methodology which suggested to investigate the ratio of lumbar motion, relative to hip motion will divide the lumbar into sections to explore movement and velocity of multi-regional of lumbar spine with relative to hip.

The additional aim of this study was to explore the relationship between the kinematic profiles of flexion and three sagittally dominant functional tasks (lifting, stand-to-sit and sit-to-stand). Exploring the relationship between the kinematic profiles of flexion and three dominant functional tasks is a principle aim of this thesis work. The kinematic profile for the anatomical regions of upper and lower lumbar spine and hip will be used in determining correlations and differences.

This section will explain the methods used to demonstrate information in multi-regional lumbar and hip as well as the correlation of lumbar-hip kinematics between flexion and other functional tasks in multi-spinal regions of lumbar.

3.1.5.1 Participants

A total of fifty three males subjects were recruited from Cardiff University (age = 29.4 ± 6.5 years; mass = 75.3 ± 16.4 kg; height = 1.69 ± 0.15 m). Cardiff School of Engineering Ethics Committee approved this study and participants were recruited via email advertisement to staff and postgraduate students as well as oral invitation directly by researcher, meaning our cohort was a convenience-based sample. This study faced difficulty to recruit a combination of female and male subjects. All of participants in this study were males, and no female decide to take position in this study, furthermore, a number of males were agreed to participate and confirmed their participation by email but they changed their mind and refused to attach the sensors to skin at experimental lab. None of the participants had a history of spinal pain or any disorder of the cervical, thoracic or lumbar spine or the hip. Furthermore, participants were free from any neurological conditions, vestibular disturbances, inflammatory joint disease or a history of spinal surgery. This study was approved by Cardiff School of Engineering Ethics Committee. Participants were recruited via email advertisement to staff and postgraduate students, meaning our cohort was a convenience-based sample. All participants provided informed, written consent. SEM

3.1.5.2 Instrumentation

Data describing lumbar spine and hip kinematics were collected using four sensors (triaxial accelerometers) with a footprint of 24 mm2 (THETAMetrix, Waterlooville, UK). Sensors were placed using double-sided tape over the spinous processes of S1, L3, T12 and the lateral aspect of the right thigh, mid-way between the lateral epicondyle and greater trochanter on the iliotibial band (ITB) (Figure 3.6.1). Each accelerometer provided axial acceleration data pertaining to absolute orientation (tilt), with respect to gravity.

Sensors were wired together in a 'daisy chain' arrangement and connected to a PC, running data collection software via USB. Data were captured at 30Hz and stored for retrospective processing. This system has been previously shown to have been an excellent repeated measure in terms of reliability relating to spinal motion analysis, with ICC ranging from 0.88-0.99 and standard errors ranging 0.4 – 5.2° (Alqhtani et al. 2015a). The accuracy of such a system has been established in a preliminary study and has shown to offer RMSEs of 0.70%-1.39% compared to a precision angle measurement table (THETAmetrix, Waterlooville, UK), when the two systems have operated to measure the movement of the axis through ±180°. These results have also been published in Alqhtani et al. (2015a).

3.1.5.3 Procedure

For measuring lumbar spine movement, authors have tended to place one sensor or marker on the spinous process of L1 and L5 (Dolan and Adams 1993; Williams et al. 2010; Ha et al. 2013; Williams et al. 2013). Other authors placed them on L1 and sacral (Lee and Wong 2002; Wong and Lee 2004; Shum et al. 2005a; Shum et al. 2010a; Shum et al. 2007a; Tafazzol et al. 2014). Different landmarks also suggested the spinous process of T12-L1 and below S2 (Esola et al. 1996) or T12 and S1 (Burdett et al. 1986; McClure et al. 1997; Ng et al. 2001; Mannion et al. 2004; Kellis et al. 2008). Recently, few studies have suggested different landmarks to classify the lumbar spine into two regions. They have used either three sensors or markers on the spinous processes of L1, L3 and S1 (Williams et al. 2010; Williams et al. 2012), L1, L3 and L5 (Ebert et al. 2014; Leardini et al. 2011) or T12, L3 and S2 (Mitchell et al. 2008; Parkinson et al. 2013). In present study, four sensors (Figure 3.6.1) were placed firmly on the skin using double-sided hypoallergenic tape over the spinous processes of T12, L3 and S1 and lateral aspect of the right thigh midway between the lateral epicondyle and greater trochanter on the iliotibial band (ITB) (Alqhtani et al. 2015). The participants' height and weight were determined prior to sensor attachment. They completed a warm-up exercise, which included flexion, extension and rotation of the trunk and sensor familiarisation, to ensure the participant became accustomed to moving with the sensors attached.

Prior to starting the actual trial, participants were asked to do one trial to familiarise themselves with the experimental procedure. Each participant stood barefoot on assigned markers and focused on a wall marker, set at a height of two metres, with arms relaxed by their side. Participants were asked to complete forward bending, backward bending, lifting an object (wooden box with handles weighing 3 kg) from the floor and to return to a standing position, moving from stand to sit on a stool and then returning to standing. No further instructions on how to move were provided.

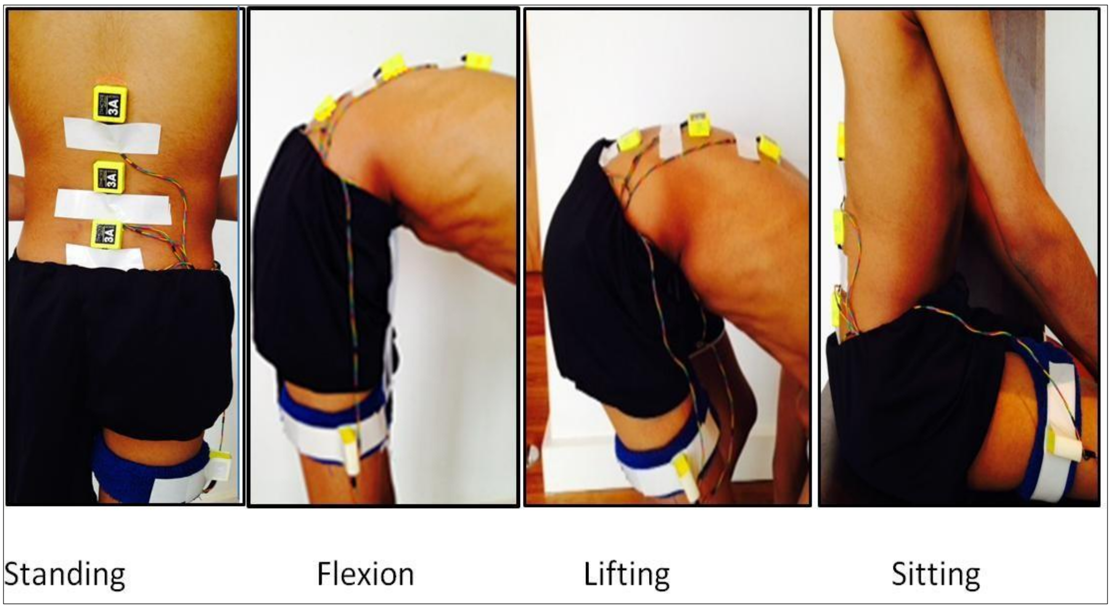


Figure 3.6.1: Schematic represents the location of three sensors on spinous processes of T12, L3 and S1 and on the lateral aspect of the thigh midway between the lateral epicondyle and greater trochanter on the iliotibial band (ITB).

3.1.5.4 Statistical analysis

Data were captured at 30Hz and the raw data were transferred to MATLAB (R2011a) Sagittal plane absolute angles for each sensor were determined, with respect to gravity, and regional range of motion was defined as the relative motion between adjacent distal and proximal sensors (relative angles). The whole lumbar (WL) spine was defined as the relative angle between the S1 and T12 sensors. The upper lumbar spine (UL) was defined as the relative angle between the L3 and T12 sensors, and lower lumbar spine (LL) as the relative angle between the S1 and L3 sensors. Hip kinematics were derived from the relative angle between the S1 and thigh sensors. Having collected the data of absolute orientation, defined as Euler angles, this was changed into rotation matrices. The resultant angles between two adjacent sensors were then computed through matrix multiplication to determine the motion of each individual spinal segment through a custom-written code in Matlab (Lee & Fung, 2003; Williams et al., 2013). The ROM data for each movement were determined and filtered at 6Hz (low-pass, Butterworth) to remove high frequency noise (Scholz et al. 2001). The ROM data were differentiated to yield the velocity. Matlab codes to run a five-point differentiation to yield angular velocity has been used in current study. These codes have been wrote and used at study by (Williams et al., (2013) for angular velocity measurement. Positive and negative velocity of the upper spine, lower spine and hip were obtained for all tasks by differentiating the range of motion data. All data were normally distributed. As the WL spine consists of six spinal joints and each of UL and LL consist three joints, therefore, WL region was normalised per segment (WL/6), while normalised per segment (UL/3 and LL/3). This normalisation enabled comparisons between the regions to be made. The kinematics of range of motion were determined as relative angle across time and angular velocity was calculated by applying 5-point differentiation of the range of motion-time data (Williams et al. 2013). The ratios of lumbar-to-hip motion for each region (UL, LL and WL) were determined for each task. Therefore, the dependent variables for this study were range of motion, peak velocity (negative and positive) and lumbar-hip ratio.

An ANOVA (One-way analysis of variance) was used to test for differences between the WL, UL and LL (SPSS ver. 20). Post-hoc analysis was applied using the Tukey procedure to determine the location of any differences. Statistical significance was accepted at a 5% level for all tests. Correlations between tasks were explored comparing range of motion and velocity profiles using Pearson's correlation coefficient calculated in a matrix laboratory (Matlab-R2011a).

3.1.6 Development of Matlab programmes

MatLab software codes were written to develop graphical figures which reflect the movement behaviours and velocity in real-time. Analysing spinal and hip data by MatLab allows the user and reader to watch and analyse the real-time graphical representation of the spinal motion as the motion is performed. Analysing large amounts of information requires a sophisticated programme such as MatLab. Therefore, a series of MatLab codes were written to obtain the tilt angles relative to gravity (absolute angles) and regional range of motion was defined as the relative motion between adjacent distal and proximal sensors (i.e. relative angles). The following flowchart provides a series of processes which used for creating the MatLab codes for each task. the aim was to explore a range of motion and velocity magnitudes

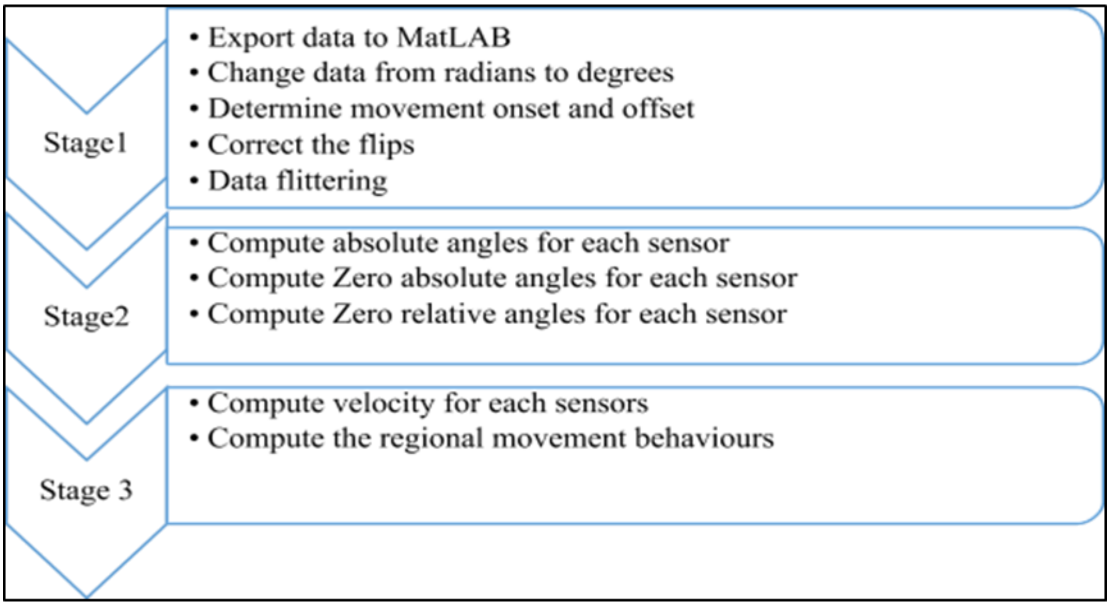


Figure 3.6.2: Flowchart illustrates the phases of writing up MatLab codes.

Codes of flexion, extension, lifting, stand-to-sit and sit-to stand tasks have written in detail for each movement in (appendix C). Each movement task has specific programme codes which are written to fit the nature and period of each particular movement. Figures 3.6.2 and 3.6.3 show the MatLab windows displaying real-time graphical representation of motion and velocity of hip during flexion, extension, lifting stand-to-sit and sit-to-stand as well as hip, upper and lower lumbar spine during flexion movement. These figures displaying the sensors movement and offer the graphical data in absolute, zero absolute and relative degrees.

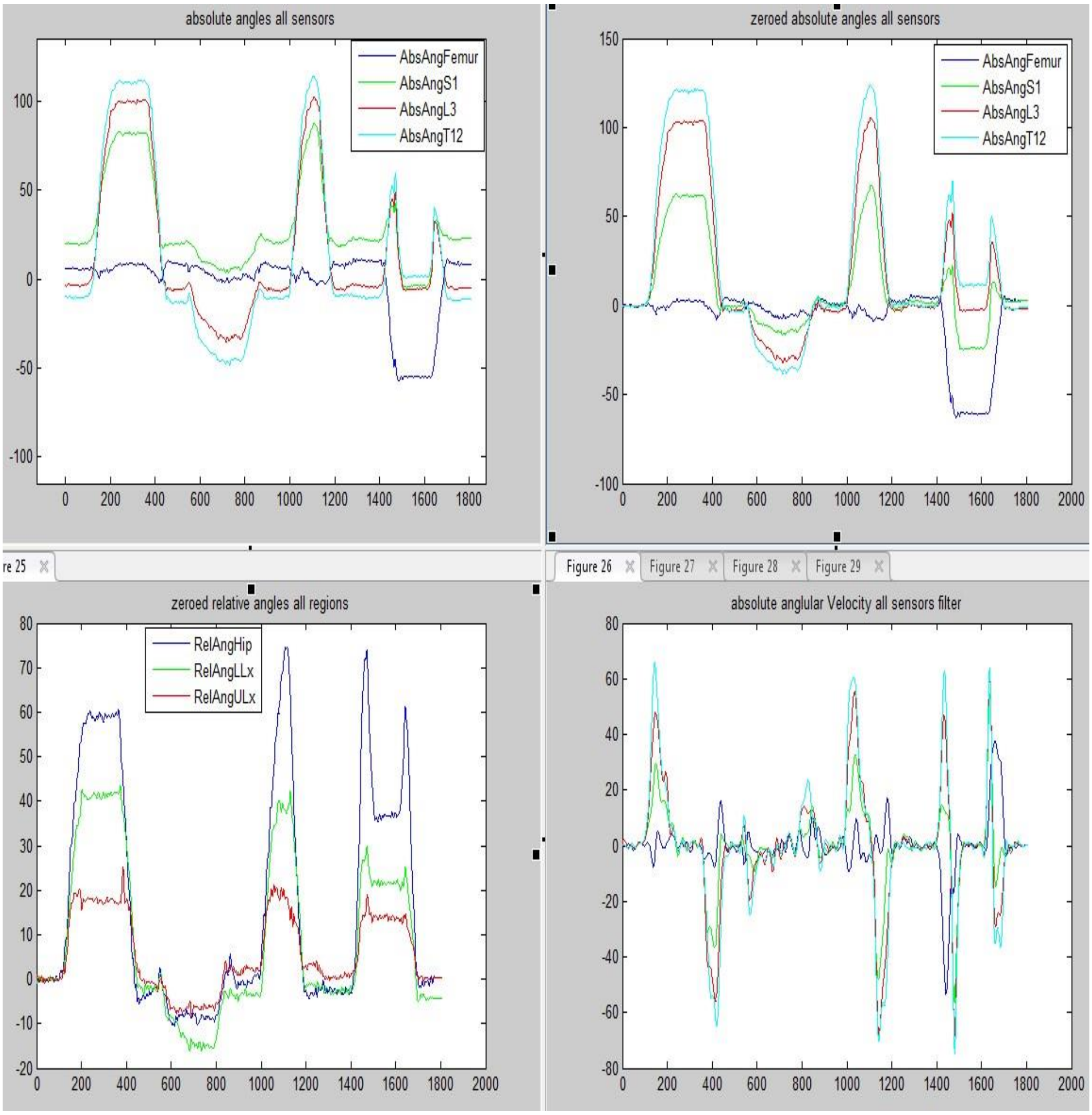


Figure 3.6.3: MatLab window displaying real-time graphical representation of motion and velocity of three sensors on spinous processes of T12, L3 and S1 and on the lateral aspect of the thigh during flexion, extension, lifting stand-to-sit and sit-to-stand.

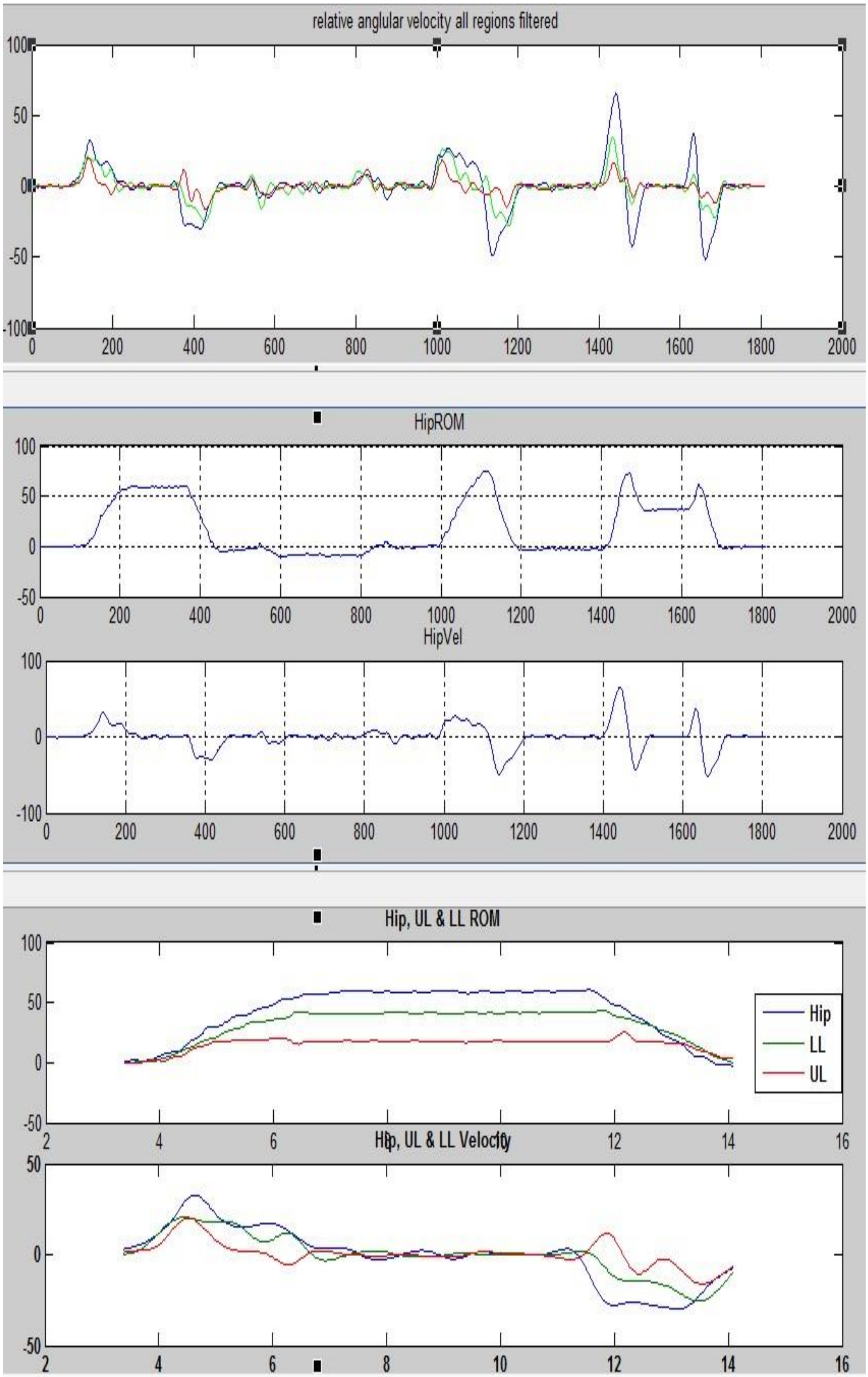


Figure 3.6.4: MatLab window displaying real-time graphical representation of motion and velocity of hip during flexion, extension, lifting stand-to-sit and sit-to-stand as well as hip, upper and lower lumbar spine during flexion movement.

Chapter 4: Results

4 Results

4.1 Correlation and RMSE

The results are summarised in table (4.1.1) with the RMSE and correlations between two devices. This experiment provided the information which explores the validity of 3A system against RT. Strong correlation between Tri-axial accelerometer sensors and Rolly Table across all measurements ranged between 0.996 and 0.999 (Table 4.1.1). Small RMSEs were shown across all tests; however, they were about 5° and 4° at roll axis when sensors were placed in cross-talk position (pitch locked at 30° and 60°). The percentages of these errors were only 1.39% and 1.33%, respectively. Figures from 4.1.1 to 4.1.6 are displaying the correlation between two systems over time when they capturing data. Angles were captured in roll, pitch axes from 0˚ to ±180˚ and, roll from 0˚ to ±180˚ when pitch locked at 30 degrees and then at 60 degrees to measure coss-tack of axes as well as measuring pitch when roll axis locked at 30 and 60 degrees slowly rotates in pitch through ±180˚.

Table 4.1.1: Root mean square error (%) and correlation between 3A system and RT with 95% confidence interval (CI) values.

Test	**RMSE (%)**	**Correlation r (95% CI)**
Roll	3.87º (1.07%)	0.998 (.997-.999)
Pitch	3.63º (1.00%)	0.999 (.996-.999)
Roll (Pitch locked at 30º)	4.28º (1.33%)	0.996 (.994-.999)
Roll (Pitch locked at 60º)	5.01º (1.39%)	0.997 (.993-.998)
Pitch (Roll locked at 30º)	3.29º (0.91%)	0.999 (.998-.999)
Pitch (Roll locked at 60º)	2.54º (0.70%)	0.998 (.996-.998)
Average	3.22 (0.89%)	0.998 (.993-.999)

❖ RMSE, root mean square error; (%) = RMSE/ (360)*100.

It is apparent from this table that the correlation between Tri-axial accelerometer sensors and Roll Table is strong overall tests with average (r=0.998). RMSE findings in table 4.1.1 ranged from 2.54º (0.70%) to 5.01º (1.39%). Interestingly, small error showed at pitch axis when roll axis was locking at 60º which more than roll and pitch axes when other axes locked at 0, while the high value of error showed roll axis when pitch axis was locking at 60º.

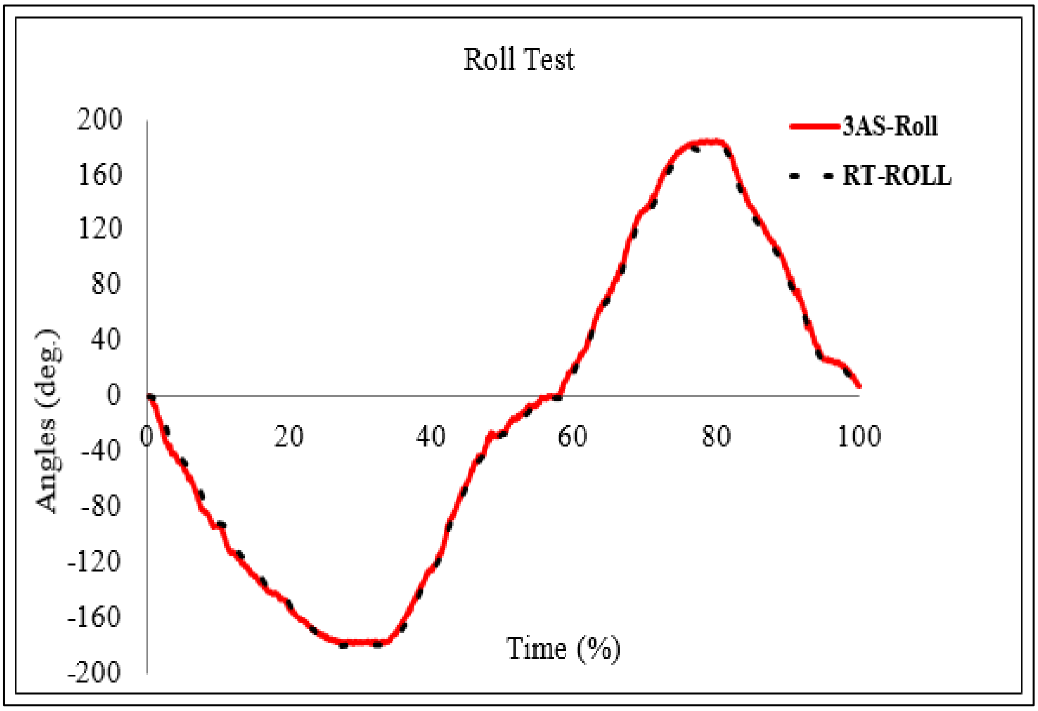


Figure 4.1.1: Roll axis test from 0˚ to ±180˚, the black dashed line represents the RT and the red solid line represents the 3A system data when the jig slowly rotates in roll through ±180˚.

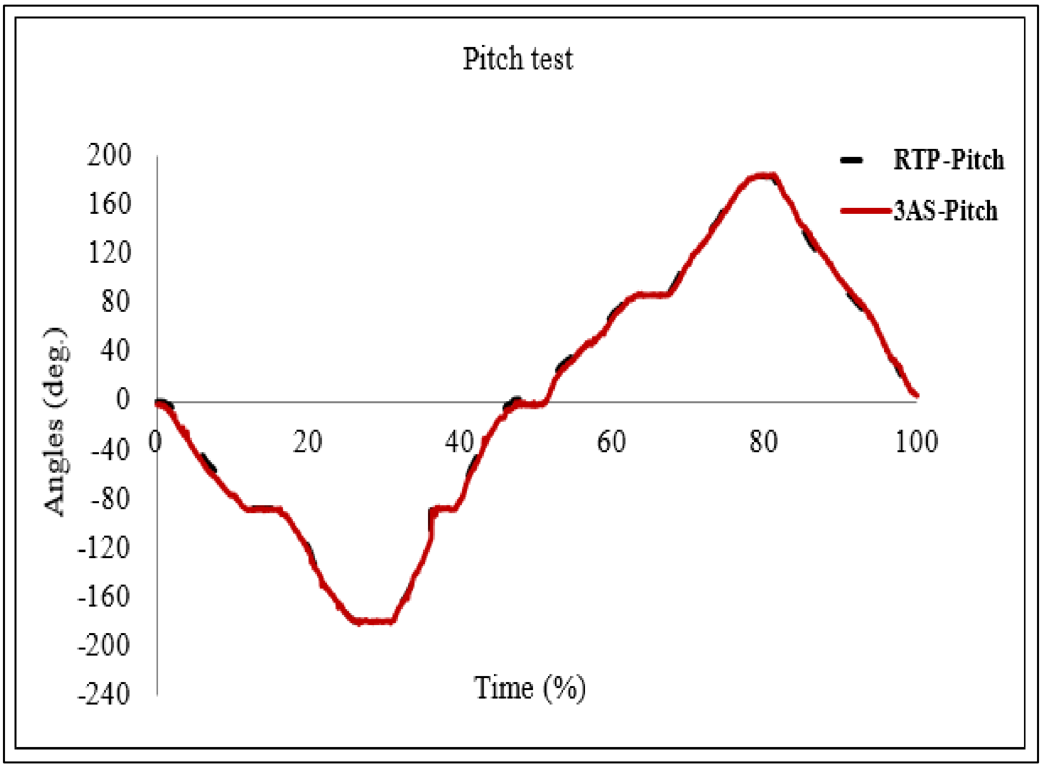


Figure 4.1.2: Pitch axis test from 0˚ to ±180˚, the black dashed line represents the RT data and the red solid line represents the 3A data when the jig slowly rotates in pitch through ±180˚.

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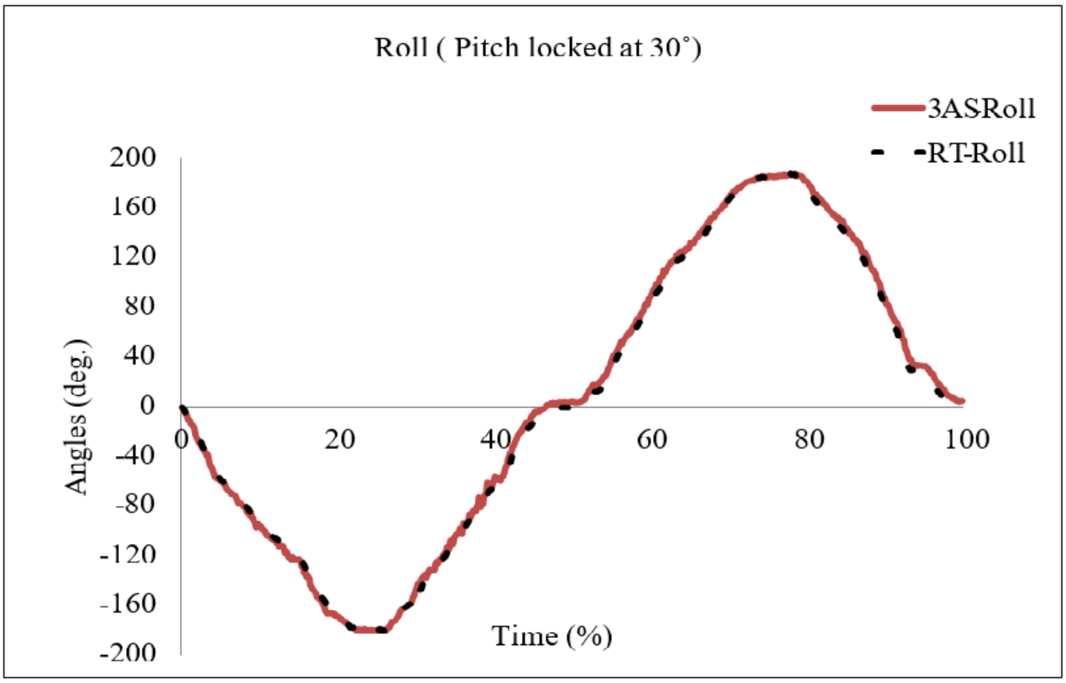

Figure 4.1.3: Crosstalk trial of Roll axis when Pitch axis locking at 30˚; the black dashed line represents the RT data and the red solid line represents the 3A system data of roll axis when the jig slowly rotates in roll through ±180˚.

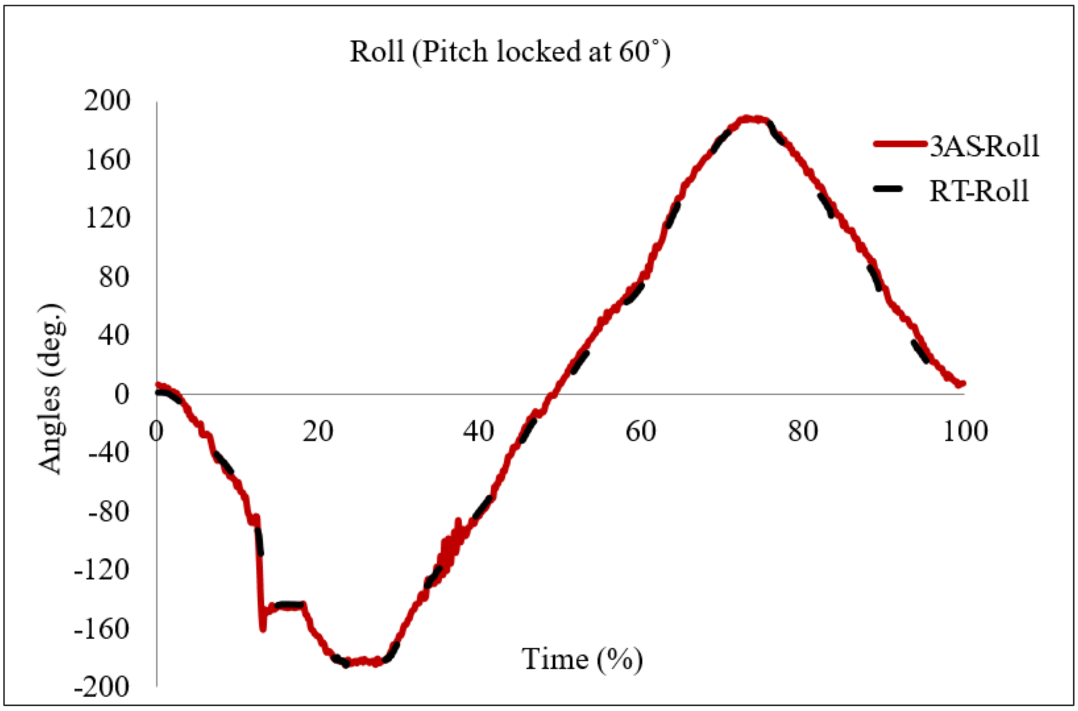

Figure 4.1.4: Crosstalk trial of Roll axis when Pitch axis locking at 60˚; the black dashed line represents the RT data and the red solid line represents the 3A system data of roll axis when the jig slowly rotates in roll through ±180˚.

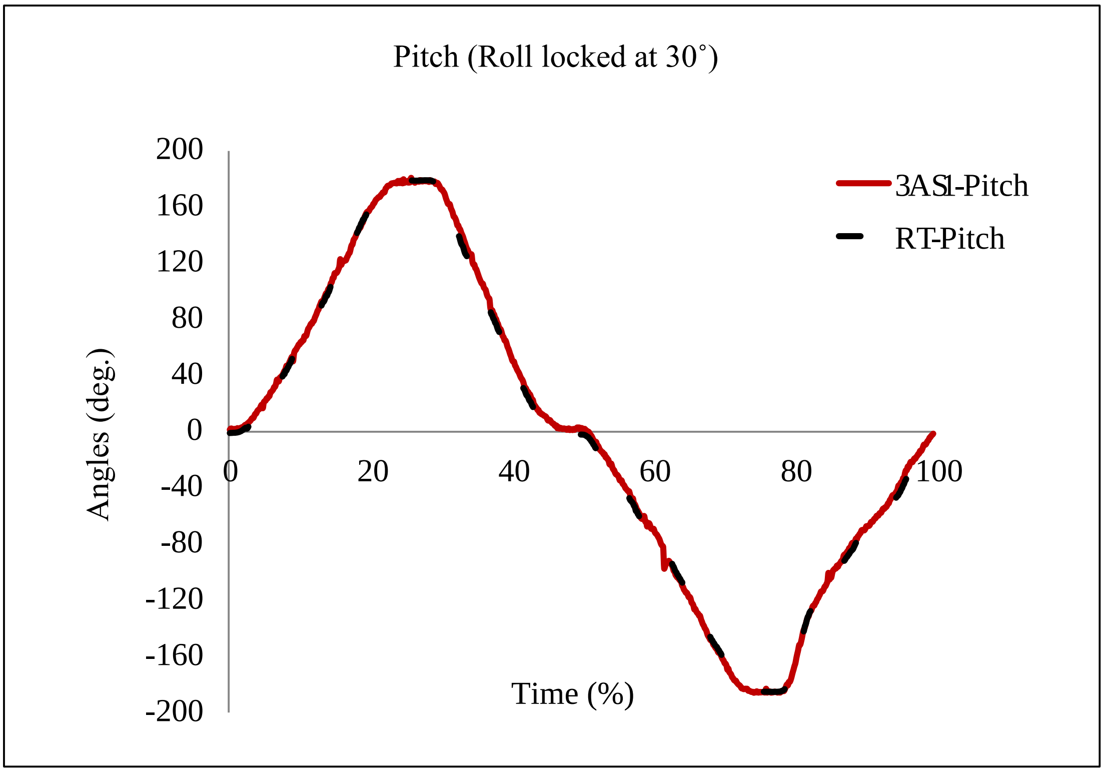

Figure 4.1.5: Crosstalk trial of Pitch axis when Roll axis locking at 30˚; the black dashed line represents the RT table data and the red solid line represents the 3A system data of pitch axis when the jig slowly rotates in pitch through ±180˚.

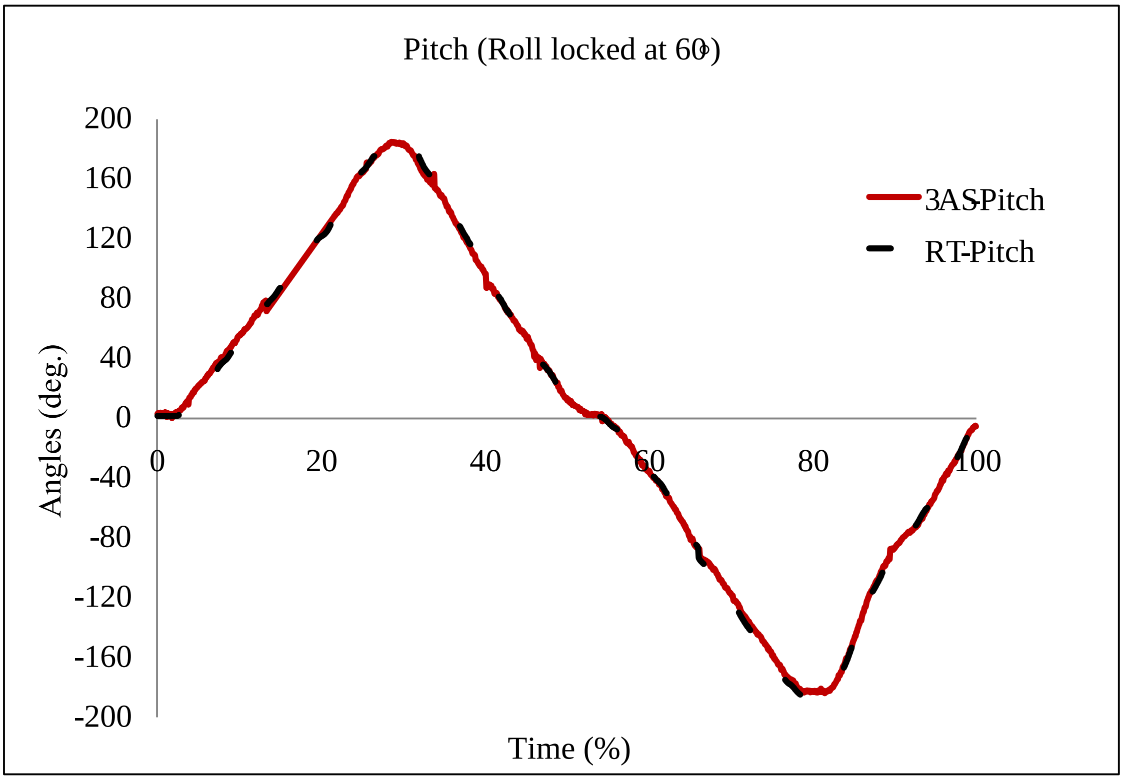

Figure 4.1.6: Crosstalk trial Pitch axis when Roll axis locking at 60˚, the dashed line represents the RT data and the solid line represents the 3A system data of roll axis when the jig slowly rotates in pitch through ±180˚

Figures 4.1.1 and 4.1.2 illustrated the strong correlation between data captured by triaxial accelerometers sensor and RT table system during roll and pitch tests from 0˚ to ±180˚. In addition, figures 4.1.3 and 4.1.4 show strong correlation between two devices when measuring the crosstalk trial of Roll axis when Pitch axis locking at 30º and at 60º and the jig slowly rotates in roll through ±180˚. Similarly, figures 4.1.5 and 4.1.6 show strong correlation between two devices when measuring the crosstalk trial of Pitch axis when Roll axis locking at 30º and at 60º and the jig slowly rotates in pitch axis through ±180º.

The findings suggest that tri-axial accelerometer sensors is a valid system and capable of measuring spinal movement in clinical settings.

4.2 Results of reliability of an accelerometer-based system in quantifying multi-regional ROM

4.2.1 Demography

The participants completed all experimental protocols without any drop out. The age, height and weight of subjects are summarized in the Table 4.2.1.

Table 4.2.1 General characteristics of subjects (N=18)

Participants (N=18)	Minimum	Maximum	Mean	Std. Deviation
Age (years)	20	43	30.6	7.6
Weight (kg)	65	117	76.6	14.4
Height (cm)	156	180	171	5.3

4.2.2 Reliability of 3A system

The Intra-class correlation coefficient (intra-tester reliability) for all regions was found to be high, ranging from mean score of .88 (95% CI .62-.93) at middle thoracic during left rotation (Table 4.2.3) and .99 (95% CI .99-.99) at head-cervical during left rotation(Table 4.2.2). There was no significant difference ($p > 0.05$) between the within-day, intra-tester scores over all regions during spinal primary movements. Errors between the intra-tester measures ranged from (SEM=0.4° and MDC=1.1°) to (SEM=5.2° and MDC=14°) for all movements and regions of the spine. From table 4.2.2, HC region showed small error (1°) at right lateral flexion while greater error (1.9°) was shown at HC extension movement. Error value (SEM=0.4°&MDC=1.1) at UT during flexion movement was the smallest relative to thoracolumbar regions in the overall spinal tests. Error value (SEM=5.2° & MDC=14°) at MT during left rotation movement was the greatest relative to other regions followed by UT (SEM=3.2° & MDC=8.8°) during right rotation. In general, the errors showed were relatively high over all regions during rotation movements, followed by extension, then flexion while lateral flexion movements had the smallest errors (Table 4.2.2, Table 4.2.3, Table 4.2.4). The percentage of errors of each particular region contribution was ranging from 4.7% at HC right rotation (Table 4.2.2) and 82 % at UL during extension (Table 4.2.4). The percentages of MDC showed difference at number of spinal regions (Table 4.2.2, Table 4.2.3, Table 4.2.4). Although, the SEM and MDC and MDC percentage scores at 95% CI in present study, SEM and MDC scores at 90% confidence interval indicated tolerable precision when (SEM < SD/2) and low variability (Boer and Moss, 2016). This study demonstrated acceptable errors as SEMs of spinal regions over all movements were smaller than (SD/2).

4.2.3 Contribution of Mult-regional spine

Multi-regional spine range of motion ranged from 3.9° at MT during flexion to 80.5° at HC during right rotation. It is apparent from table 4.2.5, that the HC range of motion of flexion, extension, and lateral flexion to right and to left and rotation to right and to left more than other spinal regions contribution through these directions. The contribution of the LL spine was more than UL spine in both flexion and extension movements, however, the contribution of the LL and UL was in right and left lateral flexion as well as at rotation movements. The contribution of LL was more than other region of thoracolumbar spine in flexion, extension, lateral flexion to right and to left, but smaller than MT and LT in rotation to right and to left. A small contribution was found at MT (3.5°) and then UL (3.9°) in flexion movement; however, a higher contribution was found at MT (right=34.8°, left=29.7°) and LT (right=21.4°, left=22.6°) during rotations to right and to left (Table 4.2.5). The regional breakdown of relative motion of the thoracolumbar spine demonstrates that 47% of the flexion motion takes place at the LL and 41% for extension, which represents a major contribution of all thoracolumbar regions. Lateral flexion relative motion demonstrates a more even spread of movement over the LL, UL and LT spine with each region contributing 24% to 26% of motion. The MT region demonstrated the greatest contribution to rotation motion, with 36% to 40% for left and right rotation, respectively. Relative contribution of head-cervical is displayed in figure (4.2.2) and the contribution of thoracolumbar regions is displayed in figure 4.2.3. Standard error of measurement (degrees) and minimal detectable changed (degree)for each spinal region during the six movements is displayed in (Table 4.2.2, Table 4.2.3, Table 4.2.4).

Table 4.2.2: Within-day, intra-examiner mean scores (three scores) and reliability measures of head-cervical (HC), upper thoracic (UT) in flexion, extension, lateral flexion to right and left and rotation to right and left.

	Intra-tester reliability					
	HC Flexion	**HC Extension**	**HC Right lateral flexion**	**HC Left lateral flexion**	**HC Right rotation**	**HC Left rotation**
Mean (SD)	66.4(12)	61.7 (11)	41.5 (7)	42.1 (10)	74.4 (10)	80.5 (14)
P value (p≤0.05) *	0.95	0.99	.85	.90	.90	.99
SEM (°)	1.2	1.5	1	1.4	1.7	1.4
MDC (°)	3.3	4.1	2.7	3.9	4.7	3.8
%= MDC/ total ROM (°)*100	4.9 %	6.6 %	6.5 %	9.2 %	6.3 %	4.7 %
ICC (95%CI)	.99 (.95 - .99)	.98 (.97 - .99)	.98 (.95 - .99)	.98 (.96-.99)	.97 (.93-.99)	.99 (.99 - .99)
	Intra-tester reliability					
	UT Flexion	**UT Extension**	**UT Right lateral flexion**	**UT Left lateral flexion**	**UT Right rotation**	**UT Left rotation**
Mean (SD)	3.9 (4)	7.1 (4)	6.5 (3)	5.4 (4)	-14.9 (16)	-11.3 (21)
P value (p≤0.05) *	.80	.95	.94	.62	.99	.99
SEM (°)	0.4	0.8	0.8	0.6	3.2	2.1
MDC (°)	1.1	2.2	2.2	1.6	8.8	5.8
%= MDC/ total ROM (°)*100	28.2 %	30.9 %	33.8 %	29.6 %	58.6 %	51.3 %
ICC (95%CI)	.99 (.94 - .99)	.96 (.91 - .98)	.92 (.82 - .97)	.98 (.95-.99)	.96 (.92-.98)	.99 (.98 - .99)

- One way ANOVA was applied using the descriptive procedure at 95% confidence interval for mean and p value ≤0.05 to determine the scores means and standard deviations.
- Intra-class correlation coefficients (ICC) with 95% confidence intervals (Cl) were calculated using intra-tester and test-retest reliability were assessed with a two-way mixed model with consistency, where the between-measure (rater) variance is excluded from the variance (Shrout and Fleiss 1979).
- Standard error of measurement (SEM(°)) was obtained using the formula: SEM = SD*(√1-ICC) (Denegar and Ball, 1993).
- Minimal detectable difference (MDC(°)) was calculated using the formula: MDC = 1.96*(√2)*SEM (Kropmans et al., 1999).
- Percentage of error (%) = MDC/ total ROM (°)*100.

Table 4.2.3: Within-day, intra-examiner mean scores (three scores) and reliability measures of thoracic and lumbar curvatures in flexion, extension, right and left lateral flexion and right and left rotation at middle thoracic (MT), lower thoracic (LT).

	Intra-tester reliability					
	MT Flexion	**MT Extension**	**MT Right lateral flexion**	**MT Left lateral flexion**	**MT Right rotation**	**MT Left rotation**
Mean (SD)	3.5 (4)	11.2 (8)	7.8 (2)	7.1 (3)	34.8 (18)	29.7 (18)
P value (p≤0.05) *	.99	.81	.94	.95	.99	.99
SEM (°)	0.7	1	0.6	0.5	1.8	5.2
MDC (°)	1.9	2.7	1.6	1.4	4.9	14
%= MDC/ total ROM (°)*100	54.2 %	24 %	20.5 %	19.7 %	14 %	46.6 %
ICC (95%CI)	.97 (.94 - .99)	.92 (81 - .98)	.91 (.90 - .97)	.97 (.92-.99)	.99(.97-.99)	.88 (.62 - .93)
	Intra-tester reliability					
	LT Flexion	**LT Extension**	**LT Right lateral flexion**	**LT Left lateral flexion**	**LT Right rotation**	**LT Left rotation**
Mean (SD)	15.0 (8)	7.9 (6)	12.1 (3)	12.4 (4)	21.4 (9)	22.6 (13)
P value (p≤0.05) *	.94	.87	.90	.90	.98	.95
SEM (°)	1.4	2.3	0.7	0.9	1.3	1.3
MDC (°)	3.8	6.3	1.9	2.5	3.6	3.6
%= MDC/ total ROM (°)*100	25.3 %	78.7 %	15.7 %	20 %	16.8 %	15.6 %
ICC (95%CI)	.97 (.95 - .99)	.92 (.83 - .98)	.95 (.88 - .98)	.95 (.95-.98)	.98 (.95-.99)	.99 (.96 - .99)

- ❖ One way ANOVA was applied using the descriptive procedure at 95% confidence interval for mean and p value ≤0.05 to determine the scores means and standard deviations.
- ❖ Intra-class correlation coefficients (ICC) with 95% confidence intervals (Cl) were calculated using intra-tester and test-retest reliability were assessed with a two-way mixed model with consistency, where the between-measure (rater) variance is excluded from the variance (Shrout and Fleiss 1979).
- ❖ Standard error of measurement (SEM(°)) was obtained using the formula: SEM = SD*(√1-ICC) (Denegar and Ball, 1993).
- ❖ Minimal detectable difference (MDC(°)) was calculated using the formula: MDC = 1.96*(√2)*SEM (Kropmans et al., 1999).
- ❖ Percentage of error (%) = MDC/ total ROM (°)*100.

Table 4.2.4: Within-day, intra-examiner mean scores (three scores) and reliability measures of thoracic and lumbar curvatures in flexion, extension, right and left lateral flexion and right and left rotation at upper lumbar (UL) and lower lumbar (LL).

	Intra-tester reliability					
	UL Flexion	**UL Extension**	**UL Right lateral flexion**	**UL Left lateral flexion**	**UL Right rotation**	**UL Left rotation**
Mean (SD)	19.4 (7)	5.0 (9)	12.6 (4)	11.3 (4)	6.3 (5)	5.3 (5)
P value (p≤0.05) *	.99	.97	.91	.95	.95	.72
SEM (°)	1	1.5	0.7	0.6	0.9	0.7
MDC (°)	2.7	4.1	1.9	1.6	2.5	1.9
%= MDC/ total ROM (°)*100	13.5 %	82 %	14.6 %	14.1 %	39.6 %	35.8 %
ICC (95%CI)	.98 (.99-.99)	.97 (.93-.99)	.97 (.92 - .99)	.98 (.95-.99)	.97 (.93-.99)	.98 (.96 - .99)
	Intra-tester reliability					
	LL Flexion	**LL Extension**	**LL Right lateral flexion**	**LL Left lateral flexion**	**LL Right rotation**	**LL Left rotation**
Mean (SD)	36.8 (6)	21.6 (14)	12.2 (4)	11.6 (3)	9.4 (8)	8.7 (7)
P value (p≤0.05) *	.94	.90	.74	.99	94	.93
SEM (°)	1.3	2.8	1.0	0.9	0.8	1.0
MDC (°)	3.6	7.7	2.7	2.5	2.2	2.7
%= MDC/ total ROM (°)*100	9.7 %	35 %	22 %	20.8 %	23.4 %	30 %
ICC (95%CI)	.95 (.98-98)	.96 (.91- .98)	.94 (.86 - .98)	.90 (.78-.96)	.99 (.97-.99)	.98 (.96 - .99)

- ❖ One way ANOVA was applied using the descriptive procedure at 95% confidence interval for mean and p value ≤0.05 to determine the scores means and standard deviations.
- ❖ Intra-class correlation coefficients (ICC) with 95% confidence intervals (Cl) were calculated using intra-tester and test-retest reliability were assessed with a two-way mixed model with consistency, where the between-measure (rater) variance is excluded from the variance (Shrout and Fleiss 1979).
- ❖ Standard error of measurement (SEM(°)) was obtained using the formula: SEM = SD*(√1-ICC) (Denegar and Ball, 1993).
- ❖ Minimal detectable difference (MDC(°)) was calculated using the formula: MDC = 1.96*(√2)*SEM (Kropmans et al., 1999).
- ❖ Percentage of error (%) = MDC/ total ROM (°)*100.

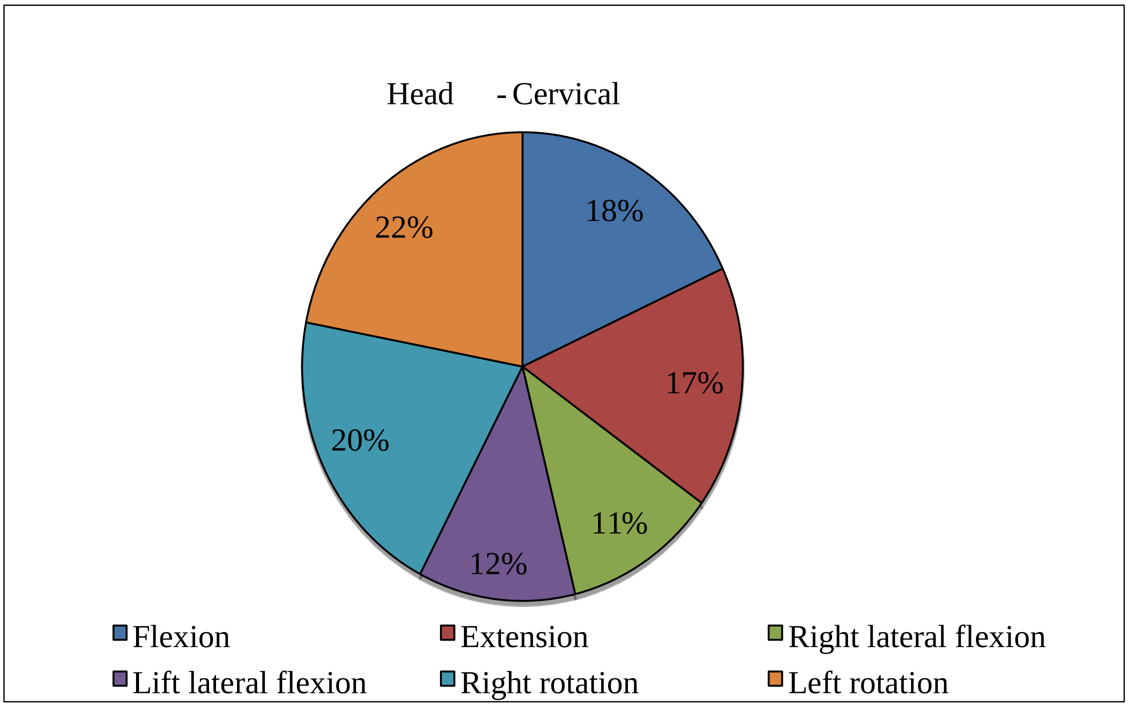

Figure 4.2.2: The percentage contribution of head-cervical during the six movements.

Figure 4.2.2 describes the percentage contribution of head-cervical during the six movements. Rotational movement to right and to lift showed the highest contribution percentages (20%, 22%), followed by flexion (18%), then extension (17%), and finally lateral flexion to right and to left (11%, 12%).

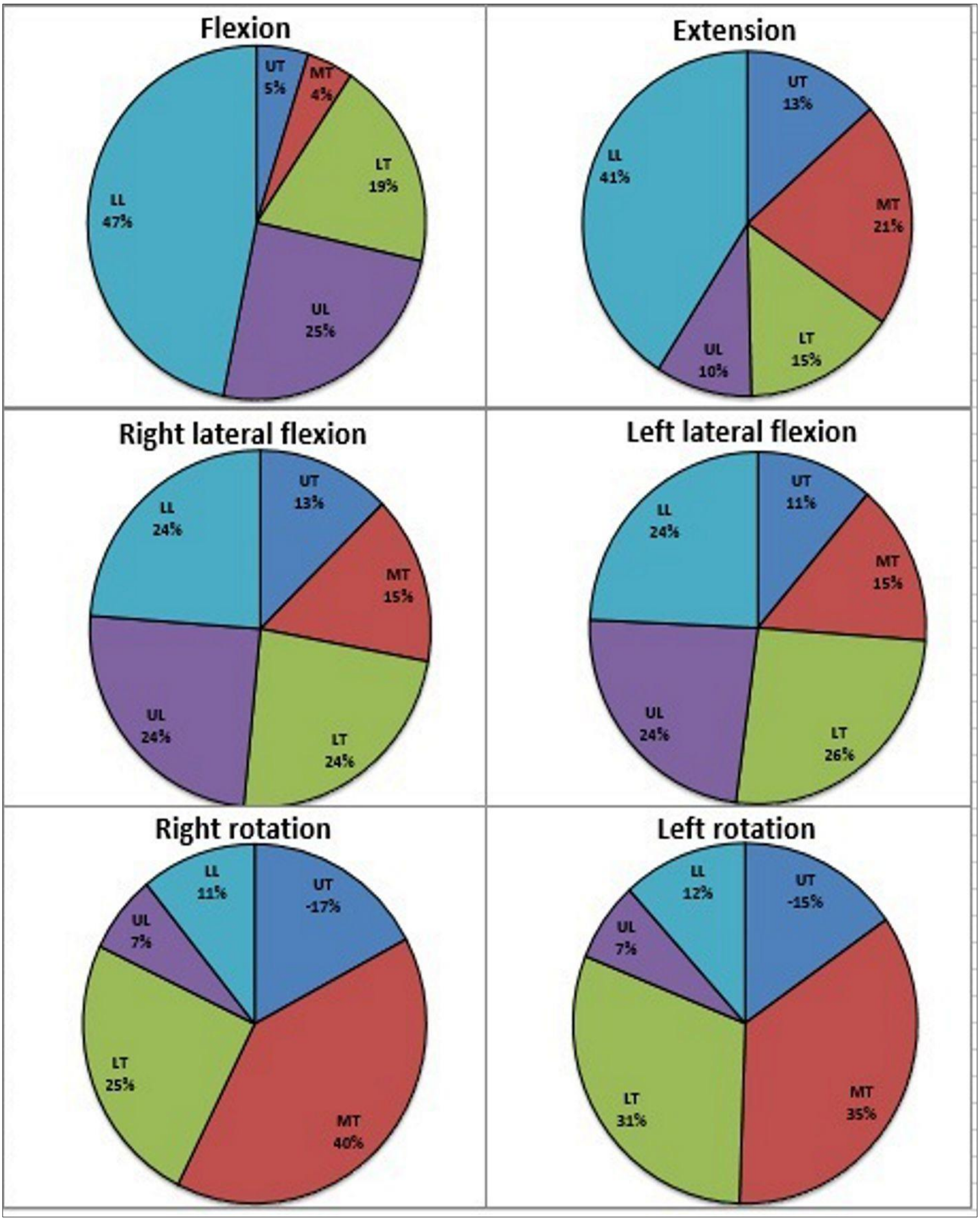

Figure 4.2.3: The percentage contribution from each spinal region during the six movements. UT: upper thoracic; MT: middle thoracic; LT: lower thoracic; UL: upper lumbar; LL: lower lumber.

Figure (4.2.3) describes the percentage contribution from five spinal regions during the six movements. In flexion movement, the contribution of the upper thoracic (5%), middle thoracic (4%), lower thoracic (19%), upper lumbar (25%) and lower lumber (47%). It appears that lower lumbar demonstrates the largest contribution while middle and upper thoracic regions demonstrate the smallest contributions. In extension movement, upper thoracic contribution (13%), middle thoracic (21%), lower thoracic (15%), upper lumbar (10%) and lower lumber (41%). Similar to flexion, lower lumbar demonstrates the largest contribution, but on the contrary upper lumbar showed the smallest contributions. In right and left lateral flexion movements, contribution of lower thoracic, upper lumbar and lower lumber are all almost similar, but at upper lumbar the smallest contribution was recorded. In right and left rotation movements, middle thoracic demonstrates the largest contribution followed by lower thoracic; but at upper lumbar the smallest contribution was recorded.

4.3 Results of the relative movement of the upper and lower lumbar spine in daily sagittal

4.3.1 Demography

The participants were completed all experimental protocols without any drop out. The age, height and weight of subjects are summarized in the Table 4.3.1.

Table 4.3.1: General characteristics of subjects (N=53)

Participants (N=53)	Minimum	Maximum	Mean	Std. Deviation
Age (years)	19	42	29.4	6.5
Weight (kg)	50	107	75.3	10.6
Height (cm)	156	186	169	1.5

4.3.2 Range of motion

The mean (SD) range of motion (normalised per segment) is presented in Table 4.3.2. Figure (4.3.1), shows the differences between lumbar spine regions (i.e. the upper, lower and whole lumbar spine regions) across different tasks (degrees). There was a significant difference in the range of motion displayed by the UL compared with the WL for flexion, lifting and sit-to-stand. Significant differences were also present between the LL and WL for flexion and lifting (Table 4.3.3). A significant difference was evident between the relative contribution from the LL and UL across all movements (Table 4.3.3), with the lower lumbar spine consistently contributing on average 63% of the total range of motion (Figure 4.3.3).

Table 4.3.2: Mean (SD) range of motion (normalised to number of segments) for the different regions of the lumbar spine and hip across different tasks (degrees)

Tasks	**WL (N=53)**	**UL (N=53)**	**LL (N=53)**	**Hip (N=53)**
Flexion	9.8 (2.4)	7.7 (3.4)	12.0 (4.4)	53.2 (14.6)
Extension	4.1 (2.6)	2.8 (3.5)	5.6 (4.3)	10 (10.7)
Lifting	9.3 (2.7)	7.2 (3.3)	11.8 (4.6)	63.2 (14.6)
Stand-to-sit	7.3 (2.8)	5.6 (3.3)	9.0 (4.9)	64.4 (17.3)
Sit-to-stand	7.3 (3.1)	5.4 (3.4)	8.9 (4.9)	64.8 (18.4)

❖ N= number of participants; WL/6 – whole lumbar spine/6; UL/3 - upper lumbar/3; LL/3 - lower lumbar/3.

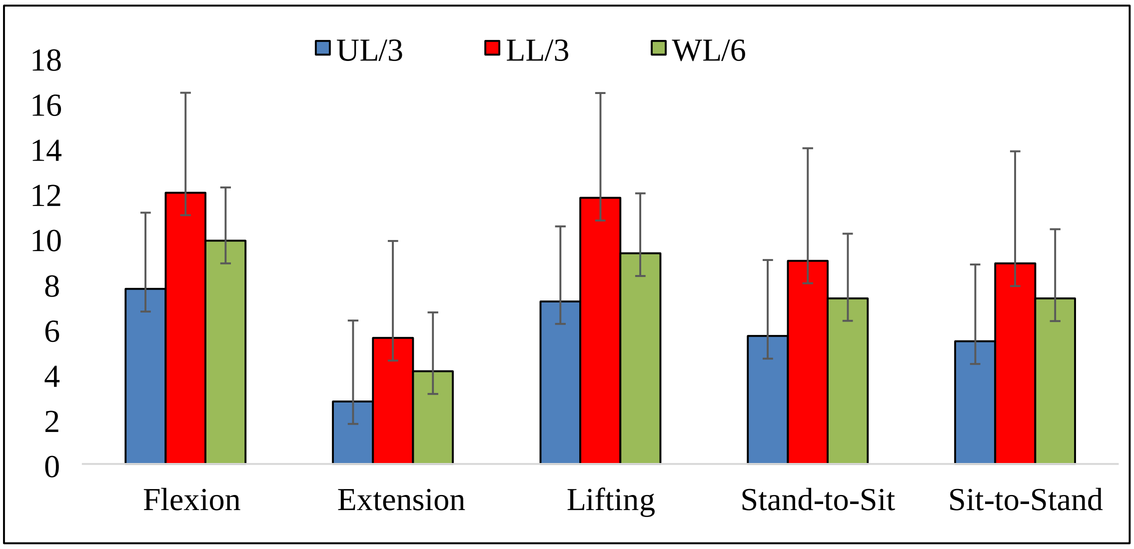

Figure 4.3.1: Mean (SD) range of motion (normalised to number of segments) for the upper, lower and whole lumbar spine regions across different tasks (degrees).

Table 4.3.3: Results of significance testing (p-value) for ROM between the different spinal regions across each task.

Difference between regional segments ROM	Flexion (N=53)	Extension (N=53)	Lifting (N=53)	Stand-to-sit (N=53)	Sit-to-stand (N=53)
UL/3 vs LL/3	<.001*	<.001*	<.001*	<.001*	<.001*
UL/3 vs WL/6	0.006*	0.191	0.009*	0.073	0.037*
LL/3 vs WL/6	0.006*	0.058	0.002*	0.073	0.109

❖ N= number of participants; WL/6 – whole lumbar spine/6; UL/3 - upper lumbar/3; LL/3 - lower lumbar/3.

❖ An ANOVA-Post-hoc analysis was applied using the Turkey procedure to determine the location of any differences and significance was accepted at a 5% level for all tests

4.3.3 Ratio

The mean (SD) peak hip-lumbar ratio per segment range of motion is displayed in table (4.3.4). A significant difference was evident between the WL-hip ratio and the LL-hip ratio for the movement of lifting only. No differences were noted for the WL-hip and UL-hip ratio. There were significant differences between the UL-hip and LL-hip ratio for all movements except extension (Table 4.3.4). Difference of mean ratio of peak (normalised) of (UL/3)/Hip, (LL/3/Hip and (WL/6)/Hip ROM (Figure 4.3.2).

Table 4.3.4: Mean (SD) ratio of peak (UL/3)/hip, (LL/3)/hip and (WL/3)/hip ROM.

Tasks	(WL/6)/Hip (N=53)	(UL/3)/Hip (N=53)	(LL/3)/Hip (N=53)
Flexion	0.20 (0.09)	0.16 (0.08)	0.25 (0.15)
Extension	0.34 (1.51)	0.20 (1.08)	0.55 (2.31)
Lifting	0.16 (0.08)	0.12 (0.07)	0.21 (0.12)
Stand-to-sit	0.13 (0.09)	0.10 (0.07)	0.16 (0.13)
Sit-to-stand	0.13 (0.08)	0.09 (0.07)	0.16 (0.13)

❖ N=number of participants; WL/6 – whole lumbar spine/6; UL/3 - upper lumbar/3; LL/3 - lower lumbar/3.

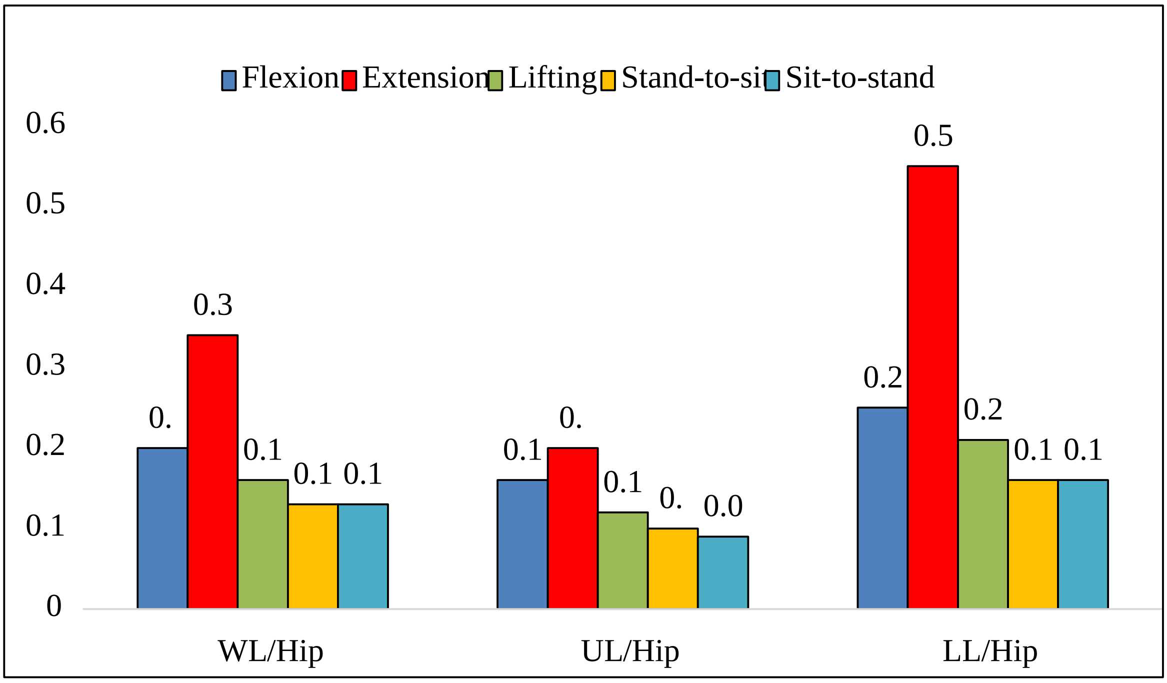

Figure 4.3.2: Mean ratio of peak (normalised) of (UL/3)/Hip, (LL/3)/Hip and (WL/6)/Hip ROM.

Table 4.3.5: Results of significance testing for ratio of peak (normalised) spine/hip ROM.

Regional segments ratio	Flexion (N=53)	Extension (N=53)	Lifting (N=53)	Stand-to sit (N=53)	Sit-to stand (N=53)
(UL/Hip vs (LL/Hip)	<.001*	0.556	<.001*	0.004*	0.002*
(WL/Hip vs (UL/Hip)	0.093	0.91	0.077	0.234	0.154
(WL/Hip vs (LL/Hip)	0.093	0.809	0.041*	0.234	0.26

❖ N= number of participants; UL/Hip: ratio of upper lumbar motion/3 relative to the hip; LL/Hip: ratio of lower lumbar motion/3 relative to the hip; WL/Hip: ratio whole lumbar motion/6 relative to the hip.

❖ An ANOVA-Post-hoc analysis was applied using the Turkey procedure to determine the location of any differences and significance was accepted at a 5% level for all t

Table 4.3.5: Results of significance testing for ratio of peak (normalised) spine/hip ROM.

Regional segments ratio	Flexion (N=53)	Extension (N=53)	Lifting (N=53)	Stand-to sit (N=53)	Sit-to stand (N=53)
(UL/Hip vs (LL/Hip)	<.001*	0.556	<.001*	0.004*	0.002*
(WL/Hip vs (UL/Hip)	0.093	0.91	0.077	0.234	0.154
(WL/Hip vs (LL/Hip)	0.093	0.809	0.041*	0.234	0.26

❖ N= number of participants; UL/Hip: ratio of upper lumbar motion/3 relative to the hip; LL/Hip: ratio of lower lumbar motion/3 relative to the hip; WL/Hip: ratio whole lumbar motion/6 relative to the hip.

❖ An ANOVA-Post-hoc analysis was applied using the Turkey procedure to determine the location of any differences and significance was accepted at a 5%

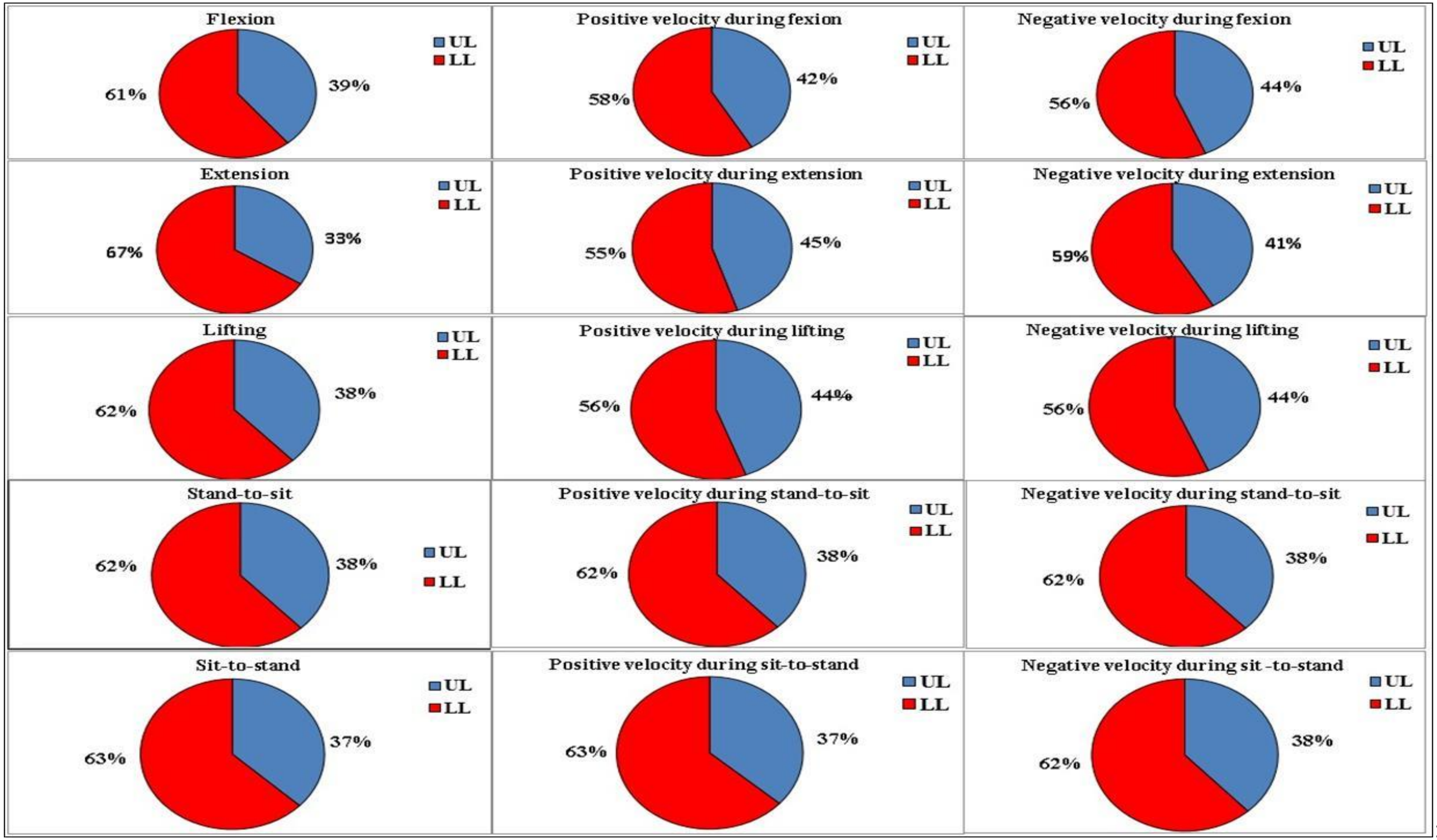

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Figure 4.3..3: The percentages of mean ROM and velocity (+ve &-ve) per-segment of upper and lower lumbar spines during five tasks.

.3.3 Velocity

Mean (SD) peak velocity for each spinal region is presented in table (4.3.6). A significant difference was evident between the WL and LL peak velocity, but only for flexion. There were significant differences between the UL and WL for peak velocity for stand-to-sit and lifting. No other tasks demonstrated 'per segment' peak velocity differences. Significant differences were determined between the UL and LL for peak velocity during all tasks, with the exception of positive velocity during extension and negative velocity during lifting. The figure (4.3.3) shows that the LL achieved greater velocity for all tasks when compared to the UL with the magnitude of difference ranging from 37% to 63%.

Table 4.3.6: Mean (SD) velocity (normalised per segment) for each spinal region across tasks (degrees/second).

Tasks	**WL (N=53)**	**UL (N=53)**	**LL (N=53)**
Flexion +ve vel	8.6 (2.8)	7.5 (2.9)	10.5 (4.7)
–ve vel	8.3 (3.4)	7.4 (3.3)	9.6 (4.5)
Extension +ve vel	5.4 (3.0)	4.9(3.0)	6.1 (4.3)
–ve vel	4.6 (2.8)	3.9 (2.9)	5.5 (4.1)
Lifting +ve vel	10.0 (3.4)	8.4 (3.9)	10.5 (4.7)
–ve vel	9.3 (3.1)	7.3 (3.3)	9.6 (4.5)
Stand-to-sit +ve vel	9.7 (3.3)	5.5 (2.5)	9.0 (4.9)
–ve vel	5.9(3.4)	3.3 (1.4)	5.4 (3.2)
Sit-to-stand +ve vel	4.3 (2.2)	3.1 (1.9)	5.5 (3.5)
–ve vel	7.5 (3.3)	5.6 (2.8)	4.2 (5.0)

❖ WL– whole lumbar spine/6; UL - upper lumbar/3; LL - lower lumbar/3.

Table 4.3.7: Results of significance testing (p-value) for velocity of UL, LL and WL segments for each task.

Difference between velocity (+ve) of regional segments	Flexion (N=53)	Extension (N=53)	Lifting (N=53)	Stand-to sit (N=53)	Sit-to-Stand (N=53)
UL vs LL	<.001*	0.228	0.021*	<.001*	<.001*
LL vs WL	0.246	0.771	0.11	<.001*	0.082
UL vs WL	0.019*	0.602	0.779	0.600	0.06
Difference between velocity (-ve) of regional segments	**Flexion (N=53)**	**Extension (N=53)**	**Lifting (N=53)**	**Stand-to sit (N=53)**	**Sit-to stand (N=53)**
UL vs LL	0.011*	0.039*	0.091	0.001*	0.001*
UL vs WL	0.421	0.535	0.029*	<.001*	0.054
LL vs WL	0.218	0.346	0.919	0.637	0.067

- ❖ N= number of participants; WL/6 – whole lumbar spine/6; UL/3 - upper lumbar/3; LL/3 - lower lumbar/3.
- ❖ An ANOVA-Post-hoc analysis was applied using the Turkey procedure to determine the location of any differences and significance was accepted at a 5% level for all tests

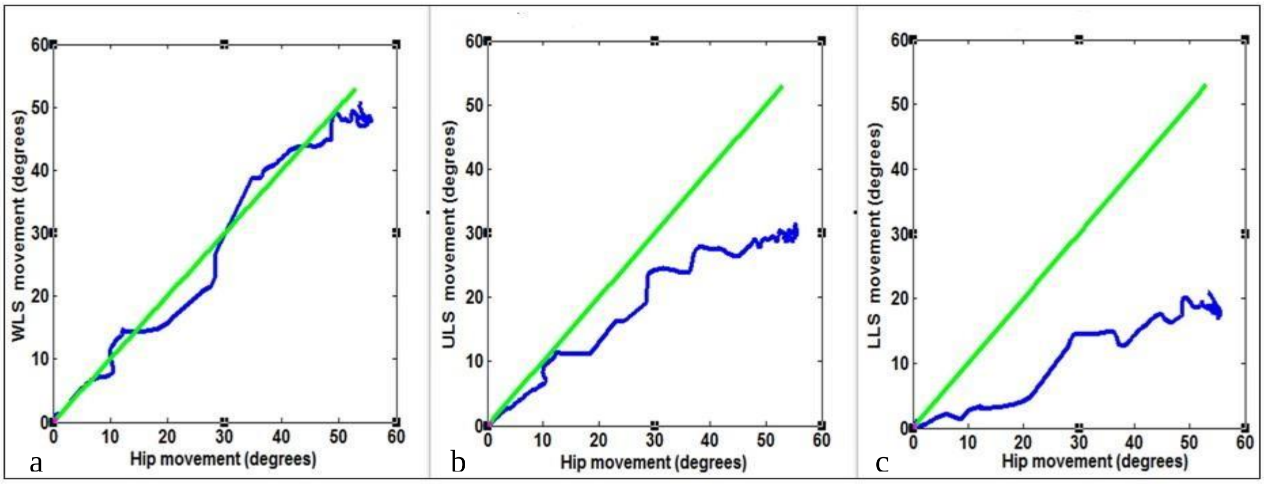

Figure 4.3.4: The phase relationship of the lumbar spine to hip movement, with the grey line representing a sustained 1:1 relationship.

Figure (4.3.4) illustrates the WL plotted against the hip and the UL-hip and LL-hip plots for comparison (the green line represents a 1:1 ratio for comparison). It shows that WL region and hip move at a similar time and rate throughout the movement phase (i.e. broadly correlating with green line), but upper and lower lumbar spine regions show a significantly greater contribution from the hip, especially in the early phase of the motion for the LL.

4.4 Results of the correlation of lumbar-hip kinematics between flexion and other functional tasks

4.4.1 Demography

The participants were completed all experimental protocols without any drop out. The age, height and weight of subjects are summarized in the Table 3.1.5.1.1.

Table 4.3.1: General characteristics of subjects (N=53)

Participants (N=53)	Minimum	Maximum	Mean	Std. Deviation
Age (years)	19	42	29.4	6.5
Weight (kg)	50	107	75.3	10.6
Height (cm)	156	186	169	1.5

4.4.2 Range of motion

Mean (sd) ROM across all tasks for each anatomical region is displayed in table (4.4.2) and a single participant's ROM-time and velocity-time graph are presented in figure 4.4.1 for the movement of flexion.

ROM-time and velocity-time graphs of hip, lower lumbar, and upper lumbar spine during flexion task of provided a clear picture for regional movement behaviour as well as the behaviour of velocities (positive and negative velocity) (Figure 4.4.1). Hip movement and velocity shown higher the two lumbar regions then followed by lower lumbar spine. Figure (4.4.1) illustrates increase of velocity at the earlier stages and then decrease at the middle stage of movement and then start to increase again at the final stage. The cycle movement started at upright standing, then full flexion and return to upright standing again.

The range of motion utilised during flexion was significantly different to that for stand-to-sit and sit-to-stand for all anatomical regions as well as differences in ROM between flexion and lifting which were observed for the hip only (Table 4.4.2).

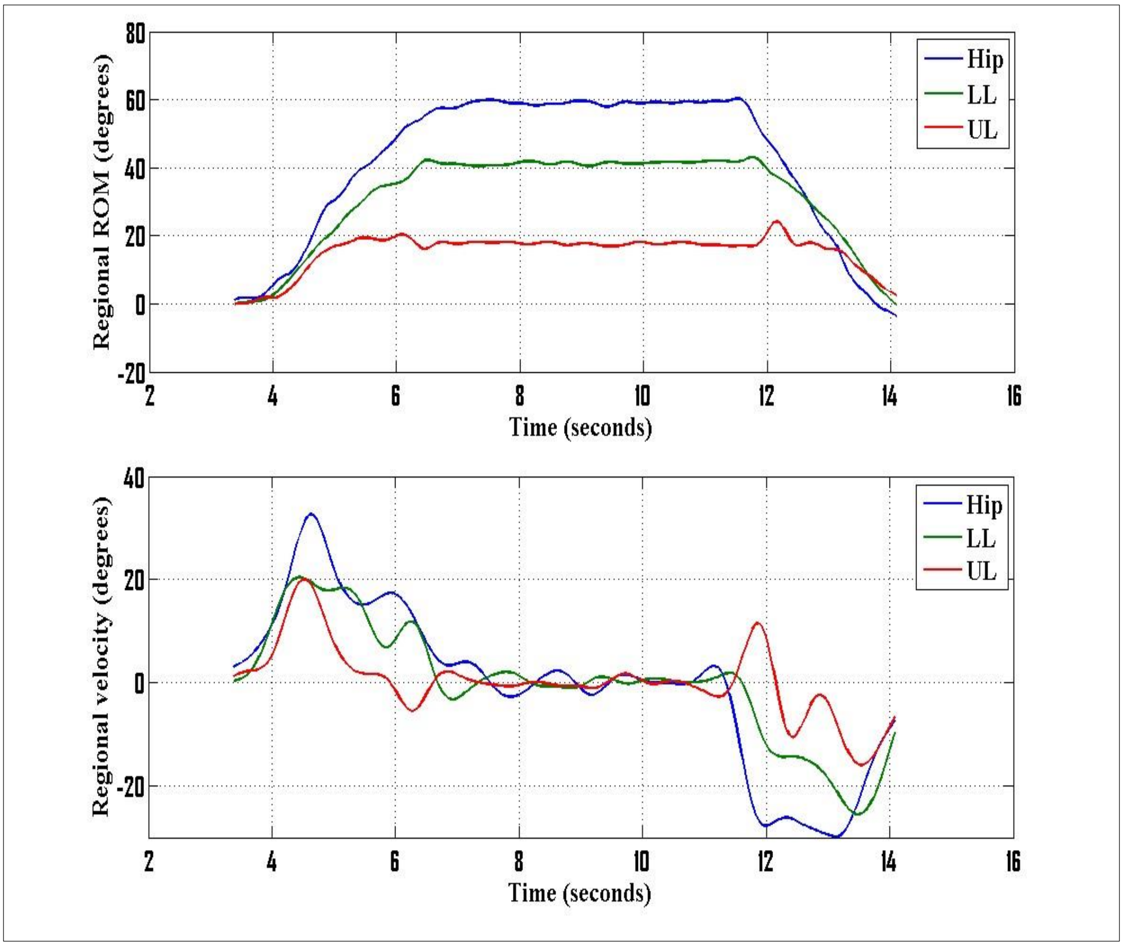

Figure 4.4.1: ROM-time and velocity-time graphs of hip, lower lumbar (LL), and upper lumbar (UL) during flexion task of individual participant.

Table 4.4.2: Mean (sd) range of motion and velocity for the four tasks and each anatomical region (UL, LL and Hip).

	ROM (°) (N=53)			Positive velocity (ºs-1) (N=53)			Negative velocity (ºs-1) (N=53)		
Tasks	**UL**	**LL**	**Hip**	**UL**	**LL**	**Hip**	**UL**	**LL**	**Hip**
Flexion	23.3 (10.1)	36.0 (13.3)	53.2 (14.6)	22.4 (8.8)	31.6 (14.1)	33.0 (18.5)	22.2 (9.9)	28.7 (13.6)	35.0 (16.9)
Lifting	21.6 (9.9)	35.4 (13.9)	63.2 (14.6)	25.2 (11.8)	35.6 (13.4)	51.5 (22.4)	23.3 (8.8)	33.4 (14.5)	50.6 (25.3)
Stand-to-sit	17.0 (10.1)	27.0 (14.9)	64.4 (17.3)	16.6 (7.7)	26.7(15.2)	57.5 (21.3)	10.0 (4.1)	16.3 (9.6)	35.0 (21.5)
Sit-to-stand	16.3 (10.2)	26.6 (14.9)	64.8 (18.4)	9.5 (5.8)	16.4 (10.6)	40.9 (22.2)	17.0 (8.6)	27.5 (15.0)	64.3 (28.4)

❖ UL, Upper Lumbar Spine; LL, Lower Lumbar Spine; ROM, range of motion.

4.4.3 Velocity

Mean (sd) velocity across all tasks for each anatomical region is displayed in table (4.4.2) and the differences between flexion and lifting velocity (positive and negative) were evident for the hip and lower lumbar spine but not for the upper lumbar spine. Differences between flexion and stand-to-sit were observed for positive and negative velocity in the upper lumbar spine, as well as differences in negative velocity in the lower lumbar spine and positive velocity for the hip (Table 4.4.2). Flexion velocity was significantly different for sit-to-stand velocity at the upper lumbar spine (positive and negative) as well as for the lower lumbar spine (positive velocity) and hip (negative velocity).

Table 4.4.3.: Demonstrating correlation *(r)* and significant differences (*p-value*) for ROM and velocity for lumbar spine and hip regions.

Regional tasks	**ROMº (N=53)**		**Positive velocity °s-1 (N=53)**		**Negative velocity °s-1 (N=53)**	
	r	***p***	***r***	***p***	***r***	***p***
UL flexion vs lifting	0.57	0.206	0.25	0.129	0.39	0.421
UL flexion vs stand-to-sit	0.52	< .001*	0.16	< .001*	0.06	< .001*
UL flexion vs sit-to-stand	0.55	< .001*	0.19	< .001*	0.03	0.007*
LL flexion vs lifting	0.83	0.545	0.29	0.084	0.53	0.017*
LL flexion vs stand-to-sit	0.7	< .001*	0.19	0.063	0.29	< .001*
LL flexion vs sit-to-stand	0.73	< .001*	0.28	< .001*	0.55	0.552
Hip flexion vs lifting	0.58	< .001*	0.47	< .001*	0.55	< .001*
Hip flexion vs stand-to-sit	0.67	< .001*	0.24	< .001*	0.31	0.999
Hip flexion vs sit-to-stand	0.66	< .001*	0.09	0.039*	0.51	< .001*

- N= number of participants; UL, Upper Lumbar Spine; LL, Lower Lumbar Spine; ROM, range of motion
- An ANOVA-Post-hoc analysis was applied using the Tukey procedure to determine the location of any differences and significance was accepted at a 5% level for all tests.
- Correlations (r) between tasks were explored comparing range of motion and velocity profiles using Pearson's correlation coefficient.

4.4.4 Correlation between tasks

Moderate to good correlations were observed between the peak of flexion during forward movement and peak of flexion during lifting for all anatomical regions investigated (0.57– 0.83). Moderate to good correlations were observed between the peak of flexion during forward movement and peak of flexion during lifting for all anatomical regions investigated (0.57– 0.83). Poor to moderate correlations were evident between peak of flexion velocity during forward movement and peak of flexion velocity during lifting task for all anatomical regions (0.25-0.55), suggesting a limited relationship between the two movements. Poor to moderate correlations were also observed between peak of flexion velocity during forward task and peak of flexion velocity during stand-to-sit and sit-to-stand (0.03-0.55), further suggesting a limited relationship between peak of flexion velocity during forward movement and peak of velocity utilised during the other functional tasks.

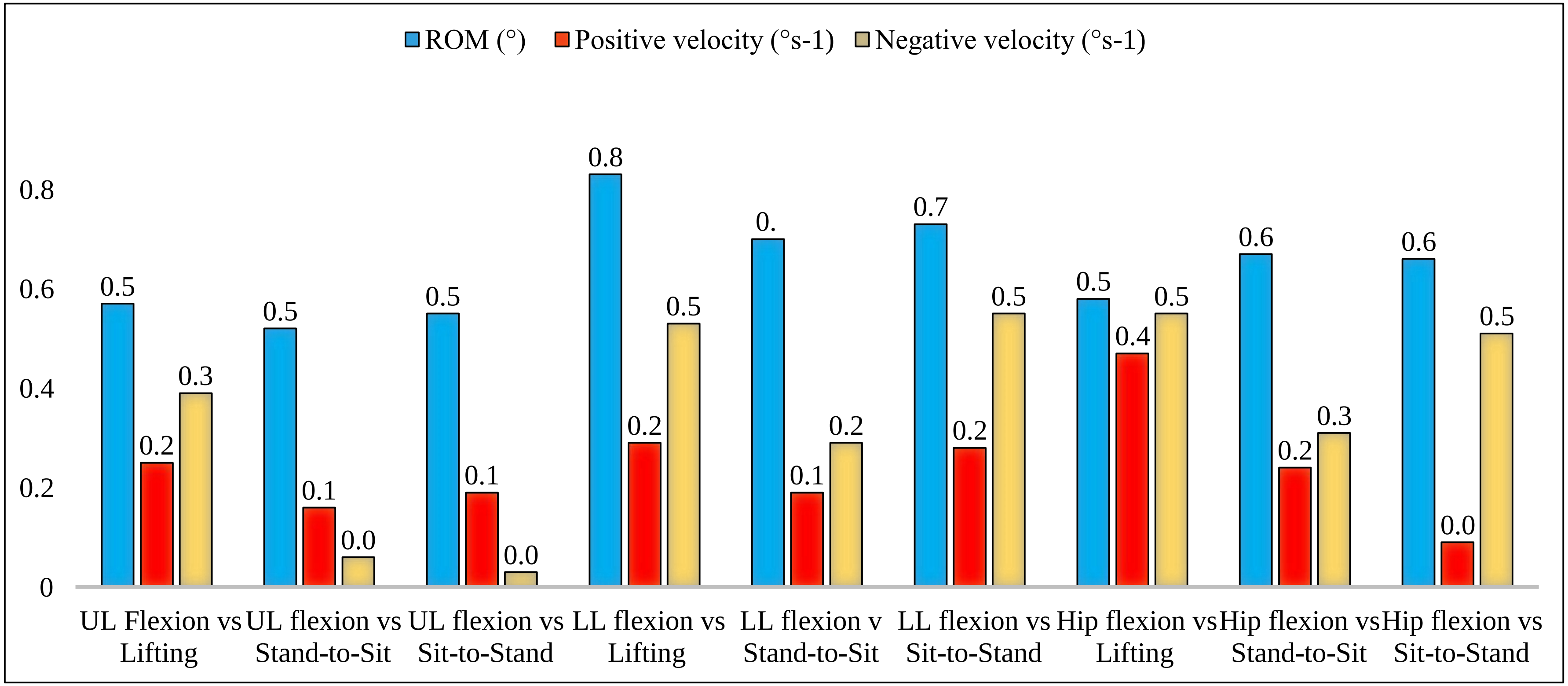

Figure 4.4.2: Relationship between the flexion and other tasks (r) at each regional range of motion and velocity.

Chapter 5: Discussion

5 Discussion

Spine and hip motion plays an essential role in daily functional activities, such as self-caring and performing occupational duties. Measurement of spinal range of motion is an important issue in clinical assessment and provides quantitative data for identifying spinal pathologies and selecting appropriate treatment and rehabilitation programs. An in-depth understanding of the physiological movement of the lumbar spine and hip could assist in developing a clinician's confidence in applying an accurate assessment and implementation of necessary treatment protocols. Measuring regional movement requires an appropriate measurement system which ideally, is capable of measuring dynamic movement in 'real time'. Additional requirements include a measurement system, which is capable of tracking and measuring movements of multi-regions, which is portable, for clinical application, valid and reliable.

Hence, a range of spinal measurement systems were reviewed and evaluated based on specific criteria (Chapter 3-section 3.1.1).

The THETAmetrix 3A Sensor Arrays system was deemed to be the most appropriate measurement system, superior to the most common systems used for measuring spine movement. Subsequently, 3A sensors has been validated against a "gold standard" rolly table to demonstrate a correlation via assessing the relative RMSEs between the two devices. All axes orientations revealed a strong correlation and the error values between the two systems (Chapter 4-section 4.1) were ranged between 1.39% and 1.33%. Reliability of 3A sensors system was also tested by measuring the range of motion of multi-spinal regions, as well as demonstrating the relative contribution of five regions from within the thoracolumbar and head-cervical regions at flexion, extension, lateral flexion to right and left and rotation to right and left.

This study was conducted when measuring primary spinal movements, three times by the same examiner, within the same day. Two protocols were applied: protocol one, consisted of sensors being placed on the forehead and T1, to measure cervical ROM; and protocol two, consisted of six sensors being placed on the spinous processes of T1, T4, T8, T12, L3 and S1 to measure thoraco-lumbar regional range of motion. The findings of this study demonstrated that ICC values for all regions were high, with errors ranging from SEM=0.4° and MDC=1.1°, to SEM=5.2° and MDC=14°, for all movements and regions of the spine. The additional movement information, gathered from multi-spinal regions, adds insight to the relative contributions to spinal movement.

Having confirmed the validity and reliability of the 3A sensors, the study establishes the sensor system as a useful tool for measuring the relative spinal kinematics.

Whilst it is common practice for clinicians to attempt to measure range of motion during an assessment of the lumbar spine, traditional single 'joint' assessment potentially provides only a limited appreciation of the wider spinal movement context and over-simplify the temporal and spatial relationships associated with the gross spinal movement. Further, multi regional analysis of motions created during daily functional tasks in the sagittal plane, such as flexion, extension, lifting, stand-to-sit and sit-to-stand, as well as the relationship between forward flexion (i.e. cardinal motion) could provide further valuable clinical data.

Kinematics of the lumbar spine and hip was measured during these tasks using three lumbar regions, relative to the hip, to investigate correlations or differences between flexion and other dominant functional tasks by measuring ROM and relative velocity magnitudes. Rather than the lumbar spine being represented as one "single" joint (S1 to T12) it was divided into two regions, the upper lumbar spine (between T12 and L3) and the lower lumbar spine (between L3 and S1) expressed relative to the hip region.

Data was collected from 53 participants, with four sensors attached to the skin over the S1, L3, T12 and lateral thigh.

The findings from the lumbar spine, viewed as a single region, was found to underestimate the contribution of the lower lumbar spine and overestimate the contribution of the upper lumbar spine. In addition, a correlation was only evident for the lower lumbar spine range of motion between lifting and flexion, while all other tasks demonstrated relatively moderate or weak correlations. The implication of this is that clinically, one should exercise caution when attempting to apply generalised findings from clinical assessment of flexion to other functional tasks, since they may misrepresent what are functionally unique kinematics.

The following subsections present the experimental study’s findings, limitations and clinical implementations within the context of the established literature.

5.1 Selection process for a spinal motion analysis system

Planning and decision making with regard to clinical intervention and treatment, partially depends on the movement quality of joint. Clinical practitioners must justify their choice of treatment modality, based on evidence which can include the objective assessment of movement (Agarwal et al.2005c). Evidence based clinical decision making therefore ideally requires the clinical observation, i.e. patient behavior etc and a valid and reproducible joint movement measurement to coincide in ‘real-time’. The first aim of the research was to assess and obtain an appropriate measurement system that was capable of tracking the movement of multi-spinal regions and once this was established ensure that the system was capable of working within the clinical setting in “real-time’.

Initially a literature based review of available spinal measurement systems was conducted to determine potential systems for a more in depth practical assessment. The review included both invasive or non-invasive systems, involving methods as diverse as optical tracking, radiology, electromagnetism, goniometry and inclinometry. Simple clinical methods, such as goniometry, inclinometry and CROM devices were quickly excluded since they are only capable of providing single point measurements in time; thus, movement behaviour across time cannot be established (Williams et al., 2013). Furthermore, measurement in three planes of motion is considered difficult, imprecise and time consuming.

Laboratory based methods were found to be poorly suited to the needs of clinical motion assessment. Opto-electronic methods have been used to measure ROM in three dimensions for the cervical spine kinematics (Edmondston et al., 2007a), thoracic spine (Edmondston et al. 2007b) and lumbar spine Ebert et al., 2014), however, whilst the systems are appropriate for research purposes, in a routine clinical assessment context, such methods are expensive and time consuming and data processing can be complex (Ha et al., 2013). Electromagnetic systems have been used to measure spinal ROM in the cervical spine (Tsang et al., 2013), thoracic spine (Hsu et al., 2008) and lumbar spine (Shum et al., 2010). However, electromagnetic systems suffer from small operating fields and subject to metallic disturbances in areas where metals are present (Ng et al., 2009; Milne et al., 1996). Inertial sensors have quantified cervical (Theobald et al., 2012) and lumbar spine ROM (Williams et al., 2013). However, spinal measurements indicate that these systems are impractical for routine clinical and research applications, due to a number of shortcomings/limitations, including the need for setting, calibration, accessibility, time-consumption, a constrained field of view and cost.

Non invasive, skin based systems were considered, however, movement artifacts were an enduring concern. Further review determined that a number of researchers have confirmed that skin based systems can provide an acceptable representation of true spinal movement (Williams et al. 2010; Ha et al. 2013; Williams et al.2013; Mitchell et al. 2008; Leardini et al. 2011; Parkinson et al. 2013).

Specific criteria were established to select an appropriate system that could capture multi regional spinal movement in ‘real time’. A number of spinal measurement systems were assessed for suitability including invasive and non-invasive systems, considering their respective strengths and limitations for setting, calibration, accessibility, time-consumption, constrained field of view and cost.

The tri-axial accelerometer sensor system recorded the highest ‘assessment criteria’ value (Table 3.2.I) and was considered to be superior to the most common systems for measuring spinal movement. The tri-axial accelerometer was selected as a result of being small, sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal motion.

Thus the 3A triaxial accelerometer based sensor system was considered to offer the greatest potential for use in physiotherapy clinics. This system was selected based on scientific criteria, however, a validation stage was required to confirm its validity against a “gold standard” reference system to assess its reliability for measuring movements which are represented during spinal movements. Whilst, this system could measure the angulation movement, velocity and acceleration precisely, this technology was limited to measuring only 2 dimensions, thus, it was required that the methodology be changed, and subsequently require the patient to be repositioned to reduce measurement errors.

5.2 The validity of the Tri-accelerometer sensors

The purpose of this study was to examine the validity of a selected system against a "gold standard'' system. The accuracy of the orientation measurements was assessed and validated against a high precision rolly "xyz" table, using yaw, pitch and roll movements. Roll and pitch axes, for the 3A system data, were examined against the rolly table axes. The roll axis was tested from 0° to +/-180° when the axes of heading and pitch were locked at 0°. The correlation between the tri axial accelerometer sensors and the rolly table was strong, the overall tests (r=0.998 at roll axis and 0.999 at pitch axis) with small percentages of error (1.07% in roll and 1.00% in pitch). It would be unusual for the spine to deviate by up to 60° during a spinal assessment through specific directions, such as the sagittal plane (i.e. flexion/extension may couple with rotation); but extra tests were undertaken when locking the roll and pitch axes, once at 30° and again at 60°. Sensors were locked once at 30° and then another test at 60° of pitch to measure the data capture in the roll axis through ±180°. A strong correlation (0.997) and percentage of error (1.33%) at 30° and (1.39%) at 60° were recorded. When locking the sensors at 30° and then another at 60° of roll, the correlation was found to be strong (0.999 and .999) and small percentages of error (0.91%) at 30° and (0.70%) at 60°. The RMSEs were 3.29º (0.91%) when rotated in pitch and locked at 30° and 2.54º (0.70%) when locked at 60° of roll. The system was found to be capable of capturing similar data through axes of roll, pitch, roll, when pitch axis was locked at 30º and 60º and pitch when roll axis was locked at 30º and 60º (Figures 4.1.1, 4.1.2, 4.1.3, 4.1.4, 4.1.5 and 4.1.6). The average of errors between two systems throughout all tests was 3.22°. The reason for the inconsistency in movement of the curve in figure 4.1.2 is that the rotary table, that was used to calibrate the 3A sensors, had to be manually rotated. Therefore, a smooth and consistent motion could not always be obtained when rotating the table. A smooth and consistent motion was particularly difficult when rotating the table at a fast (~30°/s) velocity.

The results of this study demonstrate that the average of RMSE across all tests was 3.22º (0.89%) and the average correlation between the two devices over all tests was 0.998. Comparing the findings of this study with the previous literature determined that there was strong correlation between axes values of the two systems and error values were less than those found by previous authors.

Picerno et al. (2011) investigated the accuracy of attitude and heading reference systems (AHRS) by testing the constancy of a number of system orientations, using nine inertial measurement units (IMUs) located on a Plexiglas plank. Picerno et al found that the IMUs recorded the greatest difference, calculated at 5.7 under different static situations. Lebel et al. (2015) examined AHRS systems in slow motion and found the relative precision to differ from 2° to 7°, based on the type of AHRS and the type of rotation.

The sources of error varies between studies, depending on their use of human motion. Bergmann et al. (2009) compared the anatomical joint angles of lower extremities (ankle, knee and thigh) obtained by IMUs to those calculated from positional data, from an optical tracking device, during stair ascent and found strong mean correlations (range 0.93 to 0.99), and RMSEs at 4º and 5º overall for the joint angles. Another study conducted by Favre et al. (2009) found RMSE angle errors of 8.18° in knee flexion/extension, 6.28° in abduction/adduction and 4.08° in internal/external rotation. They also reported that the RMSEs (between 4.1º and 8.11º) were moderately accurate. These findings demonstrate the level of errors to be much greater in studies, which were conducted on human beings.

Error values, similar to those found in the present study, were reported by Brennan et al. (2011) during quantification of the accuracy of inertial sensors in 3D anatomical joint angle measurements, with respect to an instrumented gimbal,. Brennan eta al found

RMSE between two instruments at 3.2° in flexion/extension, 3.4° in abduction/adduction and 2.8° in internal/external rotation. Even though the values of errors in both studies were similar, error values for the present study were less than those of Brennan et al., however, the magnitude of motion in this study (360° for each axis) was higher than the motion in Brennan et al. (2011).

Previous studies (Ferrari et al. 2010; Lebel et al. 2015; Picerno et al. 2011; Brennan et al. 2011; Sessa et al. 2012) have investigated the validity of one kind of inertial system, the AHRS, using different biomechanical procedures. However, these studies were accomplished by using different techniques and the findings varied from one experimental study to the other, making it extremely difficult to match the different findings. Furthermore, some experiments were reported to be affected by the presence of magnetic fields and motion conditions.

Clinically, there is no definitive level of acceptable error; therefore, acceptable goniometry data is typically swayable to +/- 5° error (Bruton et al. 2000) and this value could considered at large ROM joints, but not applicable for small joints such as spinal joints. To date, there is no standard value or error limit for human motion measurement systems. Therefore, this small level of error, between tri-axial accelerometer sensors and the rolly table, when compared with previous studies, was a strong indicator of its relevance to spinal biomechanical measurement. This study successfully quantified the validity of tri-axial accelerometer sensors comprising of a rolly table, which revealed a high correlation between two devices and less errors, compared with previous studies.

Thus, to summarise, for the validity study of the 3A sensor system, the second objective was to examine the validity of the 3A system against a "gold standard" system. It was hypothesised that, there will be a correlation between orientation of 3A sensors and orientation of the gold standard system. When a Pearson correlation (r) > 0.80 is reached, the null hypothesis will be rejected. Therefore, correlation r (95% CI) were ≥ 0.99

(CI=.99-.99) and then null hypothesis was rejected. The level of error, between the 3A sensors and the rolly table, was compared with previous studies, and found to be a strong indicator of its relevance to spinal biomechanical measurement. It can be concluded, therefore, that this study successfully quantified the validity of 3A sensors against a gold standard system(a rolly table), which revealed a high correlation between two devices.

5.3 Analysis of the 3A system in quantifying multi-regional spinal range of motion versus existing technologies.

This study investigated the reliability and ultimately suitability, of an accelerometer-based system for quantifying a multi-regional spinal range of motion and the relative motion of five thoracolumbar and head-cervical regions.

To evaluate the reliability, data was obtained from 18 healthy participants. The dominant six movements of the spine were measured, flexion, extension through the sagittal plane, lateral flexion to right and to left through the frontal plane and rotation to right and to left through the transverse plane. Six different regions of the spine were examined by obtaining the mean of three tests from the upper sensor, relative to mean tests and from the lower sensor for each particular region. The relative range of motion was calculated as the difference between the maximum degree of mean for the upper sensor and maximum degree of mean for the lower sensor.

The contribution of multi-spinal regions was obtained by describing the spinal range of motion when measuring the relative contribution of five regions from within the thoracolumbar region; as well as the head-cervical region during flexion, extension, lateral flexion to right and to left, and rotation to right and to left. Motion data was gathered using the reliability procedure (i.e. regional contribution, at all three planes and was repeated three times). This technique was applied to consideration of the relative contribution of HC, UT, MT, LT, UL and LL regions. The findings showed that the contribution of HC right and left rotations were 74º and 80º and flexion greater than 66º and extension 61º, respectively. The contribution of HC right and left lateral flexion was 41º and left lateral flexion was 42º.

Intra-class correlation coefficient (intra-tester reliability) for all regions was found to be high, ranging from mean score of .88 (95% CI .62-.93) at the middle thoracic during left rotation (Table 4.2.3) and .99 (95% CI .99-.99) at the head-cervical during left rotation (Table 4.2.2). There was no significant difference ($p > 0.05$) between the within-day, intra-tester scores over all regions during spinal primary movements. While, this system can be operated at high values of frequency, and is fast enough for the majority of kinematics applications, running with the operating software it can provide real-time analog data which can aid the user in monitoring and modifying movement techniques and developing positional awareness of spine curvature during dynamic movement. The results indicate that the device and methodology provided a reliable method for measuring multi- regional spinal range of motion. This is evidenced by high ICC values (i.e. > 0.88) for repeated measurements of each variable across the cohort. Indeed, the ICC values presented in this study compare favourably with other methods of spinal motion measurement, including, electromagnetic (Jasiewicz et al. 2007; Mills et al. 2007), inertial sensor (Williams et al. 2012; Theobald et al. 2012) and optoelectronic (Hidalgo et al. 2012) methods. This is the first study, however, to use a novel technique to obtain 'multi-spinal regions of the thoracolumbar spine in three planes', indicating that reliability is not compromised when measuring smaller spinal regions. In the present study, ICCs at the head-cervical region was greater than these studies, furthermore, smaller regions in the present study were found to have greater ICCs than the aforementioned studies at all movements except left rotation at MT. The present study produced ICCs ranging from .88 to .99 for smaller regions, while the aforementioned studies reported ICCs ranged from .70 to .99 for all regional movement at the cervical and lumbar spines. This provides evidence that the system is capable of providing accurate measurements for multi-spinal motion over time. Range of motion values presented in this study compare favourably with other methods of spinal motion measurement and showed a significant convergence in the results and difference with the other studies, due to different measurement systems, methods used to measure and population health conditions (Table 5.3).

Table 5.3: Comparison between the mean ROM measurements of the present study and those in previous literature.

Authors	Methods	Region	Flexion (°)	Extension (°)	Right lateral bending (°	Left lateral bending (°	Right rotation (°	Left rotation (°)
Present study	3A sensors	HC	66	61	41	42	74	80
		Thx	21	25	25	24	41	37
		Lx	55	26	24	22	15	13
Youdas et al. (1991)	CROM	HC	48	60	27	27	52	51
Lee et al. (2006)	Ultra- CMS 70p (Zebris) system)	HC	53	75.4	38	39	65	63
Sforza et al. (2002b)	Digital optoeclectric instrument (DO	HC	60	69	41	36	79	75
Lynch-Caris et al. (2008)	CROM	HC	66	85	58	57	77	74
Feipel et al. (1999b)	CA 6000 Spine Analyzer	HC	66	57	45	44	71	72
Agarwal et al. (2005)	The Spin-T goniometer	HC	57	65	44	45	70	71
Middleditch and Oliver (200	The Spin-T goniometer	HC	61	82	49	48	91	80
Theobald et al. (2012)	Inertial sensors (3DM-GX3-25)	HC	63	53	42	41	74	75
Malmstro¨m et.al. (2006)	Zebris (Zebris	HC	72	63	40	43	77	76
Average		HC	61	68	43	42	73	71
Brian D. Iveson et.al. (2010	Bubble Inclinometer	Thx+ Lx	-	-	-	-	60	58
Edmondston et al. (2007)	Optical Motion Analysis System	Thx	-	-	-	-	41	40
Hsu et al. (2008)	Flock Of Birds Electromagnic Track	Thx	26	17	26	27	42	49
Mannion et al. (2004)	A new skin-surface device	Thx	25	2	-	-	-	-
Edmondston et al. (2011)	2-D photographic image analysis	Thx	-	11				
Edmondston et al. (2012)	Radiographs and digital photographs	Thx	-	12				
Willems et al. (1996)	3 SPACE Fastrak system.	Thx	31	25	26	26	45	45
Mannion et al.(2004)	A new skin-surface device	Lx	65	14	-	-	-	-
Hsu et al. (2008)	Flock Of Birds Electromagnic Track	Lx	51	25	19	20	9	7
Van Herp et al. (2000)	Goniometry	Lx	56	22	26	26	14	12
Pearcy et al. (1985)	X-ray	Lx	51	16	17	18	4	5
Pearcy and Hindle (1989)	3 SPACE system	Lx	75	23	28	28	15	16
Hindle et al. (1990)	3 SPACE system	Lx	73	21	28	28	14	14
Russell et al. (1993)	3 SPACE system	Lx	70	25	27	27	15	15
Peach et al. (1998)	3 SPACE system	Lx	71	-	29	29	15	16
Williams et al. (2012)	Fibre-optic	Lx	37	-	-	-	-	-
Thoumie et al. (1998)	Electro-goniometer	Lx	60	-	-	-	-	-
Ha et al. (2013)	Fastrak	Lx	56	26	25	26	15	14
Ha et al. (2013)	Inertial sensors (Xsens)	Lx	56	26	26	27	16	14

5.3.1 Head -cervical contribution and reliability

The contribution of the head-cervical motion was greater at flexion (60°) than extension (61°) in the sagittal plane and greater in rotation (80°) in the transverse plane than lateral flexion (42°) in the coronal plane. The head-cervical contribution during the present study was compared with previous studies, which quantified HC movements, including using invasive techniques, such as the US-base-Zebris system (Lee et al. 2006; Malmström et al. 2006) and non-invasive techniques, such as CROM (Lynch-Caris et al. 2008), spin-T goniometer (Middleditch & Oliver 2005; Agarwal et al. 2005b), electromagnetic (Edmondston et al. 2011) and inertial sensors (Theobald et al. 2012).

The values from the present study are similar to a number of these studies; however, other studies reported varied findings, particularly in flexion and extension of HC. Some authors have reported that HC flexion is greater than extension (Feipel et al. 1999; Malmström et al. 2006; Theobald et al. 2012), while others found extension to be greater than flexion (Youdas et al. 1991; Middleditch & Oliver 2005; Agarwal et al. 2005b; Lee et al. 2006; Lynch-Caris et al. 2008). The findings of the present study are close to the average findings of past studies (Youdas et al. 1991; Feipel et al. 1999; Sforza et al. 2002; Agarwal et al. 2005c; Middleditch & Oliver 2005; Lee et al. 2006; Lynch-Caris et al. 2008; Theobald et al. 2012; Malmström et al. 2006). In these studies, the average of HC for flexion ROM was 64º and extension ROM 68º, HC right lateral flexion ROM was right 46º, left 45º, while rotation ROM at right was 77º and at left was 75º.

Using different methods, as well as different measurement systems, are likely to have contributed to the variation between the aforementioned studies. Youdas et al. (1991)

used CROM to measured cervical ROM and found right rotation of approximately 51º, however, Middleditch & Oliver (2005) found 80º at right rotation using the spine T-goniometer. The difference between the two studies was 29º, suggesting the magnitude of the differences between the techniques and devices used for measuring HC movements. In addition, the differences between these studies may be due to inter-individual factors, such as, job, lifestyle, body mass index, gender or systemic errors during measurement. Interestingly, HC left rotation was found to be greater than right rotation, contrary to a number of studies listed in table 5.3.

Intra-class correlation coefficient (intra-tester reliability) for all regions was found to be high, ranging from a mean score of .97 (95% CI .93-.99) at the head-cervical, during right rotation and .99 (95% CI .99-.99) at the head-cervical, during left rotation (Table 4.2.2). The ICCs of the present study were greater than those of previous studies, conducted by Kubas et al., 2016; Guidetti et al., 2016; Inokuchi et al., 2015; Fletcher et al., 2008; Dunleavy et al., 2015; Audette et al., 2010; Theobald et al., 2011. These studies were conducted using different systems, different designs and different populations. Inokuchi et al., (2015), Fletcher et al., (2008), Dunleavy et al., (2015) and Audette et al., (2010) measured cervical ROM by a CROM device and they used test-retest reliability, except (Fletcher et al., 2008), within day intra-rater reliability. Inokuchi et al., (2015) reported that ICCs of neck ROM measured using VICON and the CROM device were all at substantial or almost perfect levels (VICON: ICC= 0.78–0.96, the CROM device: ICC= 0.736–0.950). Fletcher et al., (2008) found ICCs ranged from 0.87 for flexion (95% CI:

0.76-0.95) to 0.94 for left rotation (95% CI: 0.87-0.97). Kubas et al., (2016) studied intra-rater and inter-rater reliability of inclinometry and Guidetti et al., (2016) also examined the intra-rater and inter-rater reliability of inclinometry and the iPhone. Kubas et al., (2016) found that the intra-rater and inter-rater reliability of the inclinometer were between moderate to excellent (ICCs= 0.53 to 0.90 and 0.69 to 0.89 respectively). Guidetti et al., (2016) found intra-rater and the interrater reliabilities were excellent (ICC=0.9) for both instruments in all movements including the rotation movements (ICC > 0.95). Repeated measures reliability for measuring spinal ROM has only been tested for the cervical spine, where an excellent coefficient of multiple correlation (CMC) and ICC values were found (0.96-0.98; 0.87-0.92), as well as small RMSE and mean absolute errors (6-7°; 3-7° for full cycle movements) (Theobald et al., 2011). However, the present study found the ICCs of all cervical movements ranged between 0.97 (95% CI: 0.93-0.99) and 0.99 (95% CI: 0.99 - 0.99), which were greater than previous studies.

The MDC statistic is useful in enabling a clinician to classify real changes from meaningless inconsistency (Quek et al., 2014). In the present study, errors between the intra-tester measurements ranged from SEM=0.4° and MDC=1.1° to SEM=5.2° and MDC=14°, for all movements and regions of the spine. From table 4.2.2, the HC region showed a small SEM (1°) at right lateral flexion, while greater SEM (1.9°) was shown at HC extension movement.

The greatest value of MDC at head-cervical flexion, extension, lateral flexion to right and to left and rotation to right and to left was approximately 4.7°. This greater value was smaller than those in other previous studies (Kubas et al., 2016; Guidetti et al., 2016;

Inokuchi et al., 2015; Kolber et al., 2013; Fletcher et al., 2008; Dunleavy et al., 2015; Audette et al., 2010) (Table 5.3.1). Even considering the variety of these studies: in design, measurement systems and populations, the MDCs of the present study were smaller than the MDCs of these studies. The percentage of MDCs for each particular movement of head-cervical region ranged from 4.7%, at HC right rotation, to 6.5% (Table 5.3.1), while for the other studies MDCs ranged from 1.1%, at cervical flexion (Dunleavy et al., 2015) and 25.4%, at cervical lateral flexion by (Kubas et al., 2016). However, Dunleavy et al., (2015) investigated the head-cervical position at forward flexion but not flexion, while Kubas et al., (2016) calculated the cervical movement to end range. Intra-tester reliability for cervical AROM measurement of persons with and without neck pain is sufficient to consider the use of the CROM in clinical practice, although clinically changes between 5° to 10° are needed to provide confidence that a real change in spine mobility has occurred (Fletcher et al., 2008). MDCs can provide evidence of age related degeneration of cervical spine motion, which in the present study was between 20 and 43 years or due to a variation of subjects heights, which ranged between 156 and 180 cm.

MDCs of the present study (i.e. within day intra-tester reliability) for cervical ROM of healthy persons ranged between 3.3° and lesser than 5°, which were less than the aforementioned studies (Table 5.3.1).

ICCs in the present study at head-cervical region was greater than previous studies. MDCs were smaller than other previous studies (Table 5.3.1) and the study design (i.e. intra-examiner within day reliability), can be considered a valuable contribution to measuring cervical movements. This system isalso useful for measuring the immediate effect of an intervention, such as head-cervical mobilisation of a patient suffering from neck pain within the same day (i.e. pre and post of treatment intervention of same day). The findings, which were produced by this system have been shown to be sensitive in detecting the differences between pre-intervention and post-intervention. Further, this newly developed method has the potential to accurately measure ROM improvement following mobilisation or following any physical approach, which is suspected to have an immediate effect.

Table 5.3.1: Comparison of the mean ROM and MDC measurements for the head-cervical region with those in the previous literature.

Author	Measurement method	Study design	Anatomical region movement	MDC (°)	MDC (%)
			Cervical flexion		
Kubas et al. (2016)	JTECH inclinometer	Intra-rater & Inter-rater reliability	48.1°	11.6°	24.10%
Guidetti et al. (2016)	iPhone®	Within day Intra-rater & Inter-rater	56.4°	7°	12.40%
	Fluid inclinometer	Within day Intra-rater & Inter-rater	56.4°	7°	12.40%
Inokuchi et al. (2015)	CROM device	Test–retest repeatability	48.1°	10.7°	22.20%
	VICON	Test–retest repeatability	48.8°	8.1°	16.70%
Audette et al. (2010)	CROM device	Within day Intrarater reliability	43°	6.3°	14.60%
Fletcher et al. (2008)	CROM device	Within day Intrarater reliability	52.9°	6.5°	12.20%
Dunleavy et al. (2015)	Optotrak	Test–retest reliability	92.4°	4.3°	4.60%
	CROM device	Test–retest reliability	90.7°	5.6°	1.10%
Present study	3Asensors	Within day Intrarater reliability	66.4°	3.3°	4.90%
			Cervical extension		
Kubas et al. (2016)	JTECH inclinometer	Intra-rater & Inter-rater reliability	73°	18.6°	25.40%
Guidetti et al. (2016)	iPhone®	Within day Intra-rater & Inter-rater	71°	7°	9.80%
	Fluid inclinometer	Within day Intra-rater & Inter-rater	72°	8°	11.10%
Inokuchi et al. (2015)	CROM device	Test–retest repeatability	57.5°	6.9°	12%
	VICON	Test–retest repeatability	54°	12.5°	23%
Audette et al. (2010)	CROM device	Test-retest reliability	68.1°	5.1°	7.40%
Fletcher et al. (2008)	CROM device	Within day Intrarater reliability	78.8°	9.3°	11.80%
Present study	3Asensors	Within day Intrarater reliability	61.7°	4.1°	6.60%
			Cervical lateral flexion		
Kubas et al. (2016)	JTECH inclinometer	Intra-rater & Inter-rater reliability	41.6°	9.8°	23.50%
Guidetti et al. (2016)	iPhone®	Within day Intra-rater & Inter-rater	50.8°	7°	13.70%
	Fluid inclinometer	Within day Intra-rater & Inter-rater	51°	7°	13.70%
Inokuchi et al. (2015)	CROM device	Test–retest repeatability	39.4°	6.1°	15.40%
	VICON	Test–retest repeatability	36°	3.6°	10%
Audette et al. (2010)	CROM device	Test-retest reliability	35°	4.2°	12%
Fletcher et al. (2008)	CROM device	Within day Intrarater reliability	41.4°	5.9°	14.20%
Present study	3Asensors	Within day Intrarater reliability	42°	2.7°	6.50%
			Cervical rotation		
Kubas et al. (2016)	JTECH inclinometer	Intra-rater & Inter-rater reliability	79°	17.1°	21.60%
Guidetti et al. (2016)	iPhone®	Within day Intra-rater & Inter-rater	72°	5°	6.90%
	Fluid inclinometer	Within day Intra-rater & Inter-rater	73°	7°	9.50%
Inokuchi et al. (2015)	CROM device	Test–retest repeatability	53°	9.3°	17.50%
	VICON	Test–retest repeatability	67°	6.9°	10.30%
Audette et al. (2010)	CROM device	Test-retest reliability	61°	6.2°	10.10%
Fletcher et al. (2008)	CROM device	Within day Intrarater reliability	74°	5.5°	7.40%
Present study	3Asensors	Within day Intrarater reliability	74.4°	4.7°	6.30%

* MDC, Minimal Detectable Change

5.3.2 Multi-thoracic regions - contribution and reliability

Quantifying a multi-regional spinal range of motion and describing its relative motion is a novel method, which is unique in enabling a regional breakdown of the range of motion within a typical clinical setting. Three regions of the thoracic spine were measured in present study, the upper thoracic (UT=3.9°, 7.1°, 6.5°,14.9°), the middle thoracic (MT=3.5°, 11.2°, 7.2° & 34°) and the lower thoracic (LT=15°, 7.9°, 12.1°& 22°) during flexion, extension, lateral flexion and rotation movements, respectively.

The contribution of the three regions of the thoracolumbar (LT, UL and LL), during lateral flexion, was very similar and greater than MT, which was greater than UT (Figure 4.2.3). The reason for limited motion in the upper and middle thoracic region could be due to the presence of the rib attachments, the sternum, the orientation of thoracic spinous process and/or facet joint's articulation. These factors may individually or in combination restrict thoracic spine kinematics; however, a further potential reason may be the thoracic curve, which is concavity in the forward direction (thoracic curve, extends from T2 to T12) and the cervical convexity extending forward (cervical convexity extends from T2 to axis vertebra) (Middleditch and Oliver 2005). Therefore, the complex formation of thoracic vertebrae, due to surrounding articulation and the vertebral curves of spine potentially limit its movement. The contribution of the UT or MT, during either flexion or lateral flexion movements was smaller than the LT contribution; however, the MT contribution, during extension and rotation was greater than the UT, while also greater than the LT.

These findings show agreement with Willems et al. (1996), except for lateral flexion. In the present study, the contribution of LT was greater than the UT and MT during lateral flexion movement contrary to the findings of Willems et al. (1996).

Various researchers have investigated thoracic regional movement, using different methods and different instruments (Willems et al. 1996; Edmondston et al. 2007; Edmondston et al. 2012; Edmondston et al. 2011), however, most have examined the thoracic region separately, without taking into account the mechanical interactions between the cervical, thoracic and lumbar motions.

Few studies (Mannion et al. 2004; Hsu et al. 2008) have tested thoracic regions relative to lumbar motion; however, their findings were closely aligned to the current study regarding all spinal movements. As a number of studies hane measured the range of motion of either thoracic or lumbar spine separately, other studies have measured the regional movement in the thoracic spine or the lumbar spine by dividing each region into smaller regions. As such, studying spinal regions in isolation may diminish the understanding of the function of the ligaments and muscles of the spine as a whole (Gill et al. 2007).

Spinal kinematics is sequential and spinal regions, i.e. cervical, thoracic and lumbar spines, are associated with and influenced by, each other. Therefore, it is suggested that if measuring each particular region separately, this could yield insufficient information to distinguish non-participation of a neighbouring region or regions. While, there is no study which has adopted such a method as used in current study, there is no option other than to sum of the three regions to compare with previous studies, which measured whole thoracic region as a single region. Using the sum of these regions (i.e. UT+MT+LT=6.5+7.2+12.1=25.8°) produced similar value to the average of studies (26.6°) which were obtained by inclinometry (Kolber et al., 2013), inertial sensors system (Bauer et al., 2016) and an iPhone (Kolber et al., 2013) at lateral flexion.

This method is currently limited by the atypical method required for measuring rotation, as a result of a need to align the motion plane with gravity. The findings showed that the contribution of MT rotation, right rotation =34.8º (40%) and left rotation= 29.7º (35%) was greater than other thoracolumbar regions, while a small contribution was found at UL (right rotation =6.3º (7%) and left rotation= 5.3º (7%). The small degree of movement, recorded at MT during flexion was 3.5 (4%), while a greater contribution was observed at LL=36.8 (47%). The limited motion in the upper and middle thoracic regions might be due to the reasons discussed above, i.e. the inflexibility of the rib cage and the thoracic and cervical curves (Middleditch and Oliver, 2005).

Intra-class correlation coefficient (intra-tester reliability) for all thoracic regions, using the 3A sensors was found to be high, ranging from a mean score of .88 (95% CI .62-.93) at the MT region at left rotation (Table 4.2.3) and .99 (95% CI .98-.99) at UT during left rotation (Table 4.2.2). There was no significant difference ($p > 0.05$) between the within-day, intra-tester scores over all regions during spinal primary movements.

Errors between the intra-tester measurements ranged from SEM=0.4° and MDC=1.1° to SEM=5.2° and MDC=14°, for all movements and regions of the spine. From table 4.2.2, the greatest value of MDC of multi-spinal regions during flexion, extension, lateral flexion to right and to left and rotation to the right and to the left was approximately 14°.

In MT, ICC was the lowest value (.88) and the MDC was the highest at the same area (14º); however, these errors were expected and are considered unavoidable. The chance of measurement errors at MT, during spinal rotation was expected, due to the position of subject (i.e. side lying position) and movement of the sensors on the spinous process of T4 was slightly altered due to the presence of relatively large amounts of soft tissue under the skin at this location. However, there was no other option, to avoid lying in the side position, due to the 3A sensor being incapable of measuring spinal rotation in the sitting or standing positions, due to requiring to be in plane with gravity. Thus, the side lying position is required for the 3A sensor to obtain the rotational movement of spine. Even, the system standardisation, which was performed before conducting rotational movement in the side-lying position to end range, was subject to artiface, due to the amount of soft tissue between the scapulae. Providing an opportunity for skin movement and a chance for measurement errors.

The movement of skin over spinous processes of vertebrae cannot be avoided when using an external skin mounted device, such as tri-axial accelerometer sensors and this error is likely to be systematic, hence leading to a relatively persistent bias in the obtained results (Gajdosik et al. 1992; Morphett et al. 2003). However, the fascia over the spinous processes is quite firmly adhered to the bone, which leads the skin movement to follow the motion of spinous processes more closely than in many other parts of the body (Lundberg 1996; Lee and Wong 2002). Error of measurement at different level of spine particular at MT could also produce a variation as a result.

Measuring thoracic rotation caused the greatest error at both right and left rotation, which is consistent with other studies and has previously been attributed to the nature of movement (i.e. task difficulty) (Ford et al. 2007) and difference between biological and flexibility aspects, across the general population (Hopkins 2000). In this instance, it may though, be due to inherently greater movement variability at this particular spinal region, or represent a slight difficultly in the ability of the clinician to fix non-moving regions, as was necessary during the measurement process of rotation.

In the context of measurement error and statistically meaningful change in the intra-examiner reliability of multi-spinal regions using the 3A sensors, no published evidence was found to which a direct comparison could be made. In the present study, the error value for UT flexion was SEM=0.4° and MDC=1.1, which was the smallest value at thoracic regions. Error values of SEM=5.2° and MDC=14°, at MT, during left rotation movement was the greatest relative to other regions; followed by UT right rotation of SEM=3.2° and MDC=8.8°. These errors were relatively high over all regions during rotation movements, followed by extension and then flexion; while for lateral flexion movements the MDC was smaller (Table 4.2.2, Table 4.2.3, Table 4.2.4). The MDC's of UT, MT and LT were nearly a third of the contribution for each region (Table 5.3.2), however, these errors were anticipated, due to the factors which are mentioned above. The SEM and MDC scores at 90% confidence interval indicated "tolerable precision" when (SEM < SD/2) and low variability (Boer and Moss, 2016). Present study SEMs were smaller than their half of standard deviations at 95% CI over all movements. Johnson et al., (2012) have examined the whole thoracic rotation from different positions using goniometry and inclinometry and found ROM reached high reliability with either a single or two trials, measured on one day and MDC ranged between 3.7º and 6.5º. The increased MDC of ROM in the present study, particularly at LT, was similar to that reported by Johnson et al., (2012). MDCs of smaller regions (i.e. LT) in the present study was similar to what was reported by Johnson et al., (2012) during whole rotation and is a strong indicator that the newly developed method has the potential to accurately measure ROM at smaller regions.

Generally, the possibility of MDC percentage increases are proportional to the smaller contributions from smaller regions, while it decreases proportionally with the greater contribution from the whole region, such as the whole thoracic spine. There is no study yet conducted that determines the MDC of UT, MT and LT in dynamic or static postures, which makes it very difficult to compare the present study with previous studies. The author has therefore attempted to derive comparisons between the current study SEMs and MDCs with previous studies, whilst acknowledging that the measurements were conducted with different: postures, movements, methods and different static analyses. Sheeran et al., (2010) found typical errors for intra-tester analyses ranged between 1.7º and 3.7º and inter-tester typical error between 2.0º and 4.7º. They reported that the measurement system (spinal wheel) demonstrated excellent 'within-day' and high 'between-day' reliability and it may be used in conjunction with a 2D camcorder to provide clinically useful visual evaluation of postures for assessment, intervention monitoring and feedback during postural re-education. They have conducted their findings in static postures (i.e. sitting position) while this study evaluated the reliability during primary movements of multi-spinal regions. It is known, that the likelyhood of errors, during dynamic movements, are greater than static positions. Errors, during this present study, were found to be smaller than static position errors, compared with those of Sheeran et al., (2010) which were conducted during sagittal plane analysis, only. The present study reported the highest value of errors at MT during rotational movement and produced dynamic data SEMs of intra-tester ‘within day’ ranged from 0.5º to 1º for thoracic regions and 0.8º for lumbar regions during flexion movement. Kobler et al., (2013) found MDCs of the whole thoracic flexion, which measured once by iPhone and once by Bubble inclinometer produced approximately 6º and 7º, respectively and thoracic lateral flexion for iPhone =4º and Bubble inclinometer =6º. Also, they found MDCs of thoraco-lumbar-pelvis flexion by an iPhone of 6º and Bubble inclinometer 7º and extension by iPhone of 9º and Bubble inclinometer 6º. In a more recent study by Bauer et al., (2016), which measured thoracic lateral flexion they found MDC was close to that measured by Kobler et al., (2013), however, both measured the whole thoracic region using ‘within day’ intra-rater and ‘test retest’ designs (Table 5.3.2). Although, these studies demonstrated greater MDCs than those found in the current study, except MT during right rotation. However, the present study is unique in that it is used for measuring more than one region at the thoracic spine and the smaller contribution of the smaller regions. The small contribution of these regions (i.e. UT, MT& LT), combined with skin movements over the spinous processes and and individuals' biological differences produced MDCs, which were large, compared to the relatively smaller regional contributions. The percentages of MDCs were greater at UT and MT regions in all directions and LT at flexion movements (Table 5.3.2). The studies of Mannion et al. 2004; and Hsu et al. 2008 tested the thoracic regions and their findings closely aligne to the present study, when the three region's contributions are summated to be one region, measuring ROM extension. The present system has similar MDCs to the other systems, or greater at some regions (i.e. smaller regions), which cannot be compared with previous studies, due to there being no study which has yet measured MDC for multi-regions at the thoracic region.

This results, produced by this system, are sensitive for detecting the differences between pre- and post-interventions, thus, this newly developed method has the potential to be used to accurately measure any improved ROM following mobilisation or following any physical approach which has immediate effect.

The results would have implications within the context of investigating the immediate effect of an intervention on thoracic ROM on patients who are suffering from back stiffness and/or pain. The present study findings could potentially be adopted to assess pre and post-surgical ROM at the thoracic region or thoraco - vertebral mobilisation which may improve ROM and function immediately.

Fletcher et al., (2008) report that currently overall improvements of between 5° to 10° are required to provide confidence that a real change in spine mobility has occurred, the smaller regions of the thoracic spine, in present study, were found to be less than 10°, except one movement (left rotation at MT). In addition to the small contribution of the smaller regions, producing MDCs, close to that found in previous studies, when measurements of whole region ROMs are considered, this system is as sensitive as existing systems for measuring whole regional ROM.

Table 5.3.2: Comparison of the mean ROM and MDC measurements for three regions of thoracic with those in previous literature.

Author	Measurement method	Study design	Anatomical region movement	MDC (°)	MDC (%)
			Thoracic regions flexion		
Present study	3Asensors	Within day intra-rater	3.9°	UT 1.1°	28.20%
		Within day intra-rater	3.5°	MT 1.9°	54.20%
		Within day intra-rater	15°	LT 3.9°	26%
			Thoracic regions extension		
Present study	3Asensors	Within day intra-rater	7.1°	UT 2.2°	30.90%
		Within day intra-rater	11.2°	MT 2.7°	24%
		Within day intra-rater	7.9°	LT 6.3°	12.60%
			Thoracic regions lateral flexion		
Present study	3Asensors	Within day intra-rater	6.5°	UT 1.6°	24.60%
		Within day intra-rater	7.2°	MT 1.4°	19.40%
		Within day intra-rater	12.1°	LT 1.9°	15.70%
			Thoracic regions rotation		
Present study	3Asensors	Within day intra-rater	14.9°	UT 8.8°	59%
		Within day intra-rater	34°	MT 14°	41.10%
		Within day intra-rater	22°	LT 3.6°	16.30%
			Thoracolumbar-Pelvis flexion		
Kolber et al. (2013)	iPhone®	Within day Intra- & Inter-examenr	107°	6°	5.60%
	Bubble inclinometer	Within day Intra- & Inter-examenr	107°	7°	6.5°
			Thoracolumbar-Pelvis extension		
Kolber et al. (2013)	iPhone®	Within day Intra- & Inter-examenr	27°	9°	33.30%
	Bubble inclinometer	Within day Intra- & Inter-examenr	28°	6°	21.40%
			Thoracolumbar lateral flexion		
Kolber et al. (2013)	iPhone®	Within day Intra- & Inter-examenr	32°	4°	12.50%
	Bubble inclinometer	Within day Intra- & Inter-examenr	30°	6°	20%
Bauer et al., (2016)	Inertial sensors systems	Intrarater-Test retest	18°	4.6°	25.50%

* UT, Upper Thoracic; MT, Middle Thoracic; LT, Lower Thoracic; MDC, Minimal Detectable Change

5.3.3 Multi-lumbar region - contribution and reliability

Findings at the two regions of the lumbar spines produced UL=19.4°, 5°, 12.2° & 5.3; LL=36.8°, 21.6°, 12° & 8.7° at flexion, extension, lateral flexion and rotation movements, respectively. The contribution of the LL spine was found to be greater than the UL spine during flexion, extension and rotation; but similar during lateral flexion. The LL spine flexion recorded a high value that was then followed by extension and lateral flexion, which was greater than the contribution by rotation. In addition, the UL spine contribution was greater than lateral flexion, while extension and rotation movement were similar (Table 4.2.2). The findings of the present study, which obtained measurements from the lumbar spine as a single joint (i.e. whole lumbar spine), found similar movement to the findings of other studies including Pearcy & Hindle (1989), Hindle et al. (1990), Russell et al. (1993) and Peach et al. (1998). These studies used different systems X-ray (Pearcy & Hindle 1989) and the 3 SPACE system for measuring the lumbar spine.

The actual range of motion values were similar to those reported previously for the thoracic spine (Mannion et al. 2004; Hsu et al. 2008) and lumbar spine (Van Herp et al. 2000; Ha et al. 2013). The findings were also similar to two smaller regions of the lumbar spine (Mitchell et al. 2008), in flexion and three smaller regions of the thoracic spine (Willems et al. 1996).

Data acquisition, describing multiple spinal regions, enables the observation of the relative contribution of each region to the overall motion; thus, clinicians can now access a wealth of information regarding spinal movement behaviour. For example, the movement of extension displays up to four times greater movement at the lower spine, compared to the upper spine. The majority of extension occurs in the mid-thoracic region of the thoracic spine, with smaller contributions occurring from above and below. Subsequently, this allows the regions of relatively altered mobility to be identified and targeted for treatment, as changes in the relative contributions to motion are likely to alter the movement and loading behaviour of specific anatomical structures (Adams et al. 1980). The contribution information from multi-regional breakdown adds insight to the relative contributions to spinal movement behaviour, which was not previously accessible.

In lumbar regions, ICCs of UL and LL spines ranged from .90 to .98 and MDCs were ranged from 1.9º to 4.1º at two regions over all movements. Right rotation of LL was found to be more reliable (ICC=.98), while the left rotation of LL was recorded as less reliable (ICC=.90). It is still unknown whether any study has yet been conducted to determine the MDC of UL and LL in dynamic or in static posture. Therefore, again there is no other option other than to compare the present study MDCs with previous studies, which were conducted by different methods and different static analyses. Even for sensors firmly adhered to the spinous processes, the variation of ICCs at both right and left rotation at similar region (LL), could produce variation due to changes in the the position from right to left. The findings of present study were found to be similar to findings from previous studies, such those of Williams et al. (2010), who investigated the LL region and lumbar spine as a whole by fibre-optic sensors. Kobler et al., (2013) found MDCs of whole lumbar flexion by iPhone of 8º and Bubble inclinometer of 9º. Also, Bauer et al., (2016) found MDCs at lumbar flexion (inertial measurement unit system (IMU) of 3.7º. Bauer et al., (2016) found MDCs at lumbar extension of 6.5º and lateral flexion of 4.6º, although, these studies demonstrated greater MDCs than those found in present study, except LL during extension. However, the contribution of the present study at LL extension was similar to the whole lumbar extension contribution in Bauer et al., (2016) as well as both studies the MDCs were similar. When the contribution and MDCs of flexion at UL and LL spine in present study were summed, the total ROM was found to be close to Kobler et al., (2013) ans Bauer et al., (2016). However, the sum of the MDC in the present study was greater than Bauer et al., (2016) but less than Kobler et al., (2013). Even the reported difference between MDCs in these studies, Kobler et al., (2013) and Bauer et al., (2016) enable clinicians and researchers to objectively quantify the lumbar movement and movement dysfunction associated with LBP treatment efficacy. The present study used MDC_{95} (%95 CI), while Kobler et al., (2013) and Bauer et al., (2016) considered MDC_{90} when interpreting their changed scores, even though, MDCs of the present study were smaller then these studies, except lumbar extension (Table 5.3.3). As discussed previously the MDCs of UL and LL spine are considered unique, therefore, MDC percentage, which is obtained at different parts of human body are used for comparison with the present study findings. Values of MDC were relative to the regional contribution of spine in the present study and those obtained by Alenezi et al., (2016) found an MDC greater than its relative regional contribution, which produced high percentage (Table 5.3.3).

Table 5.3.3: Comparison of the mean ROM and MDC measurements for two regions of lumbar spine with those in previous literature.

Author	Measurement method	Study design	Anatomical region movement	MDC (°)	MDC (%)
			Lumbar regions flexion		
Present study	3Asensors	Within day Intra-examenr	UL 19.4°	UL 2.7°	13.90%
			LL 36.8°	LL 3.6°	9.70%
			Lumbar regions extension		
Present study	3Asensors	Within day Intra-examenr	UL 5°	UL 4.1°	82%
			LL 21.6°	LL 7.7°	35.60%
			Lumbar regions lateral flexion		
Present study	3Asensors	Within day Intra-examenr	UL 12.2°	UL 1.6°	13.10%
			LL 12°	LL 2.7°	22.50%
			Lumbar regions rotation		
Present study	3Asensors	Within day Intra-examenr	UL 5.3°	UL 1.9°	35.80%
			LL 8.7°	LL 2.7°	31%
			Lumbar flexion		
Kolber et al. (2013)	iPhone®	Within day Intra- & Inter-examenr	50°	8°	16%
	Bubble inclinometer	Within day Intra- & Inter-examenr	45°	9°	20%
Bauer et al., (2016)	Inertial measurement unit (IMU)-systems	Intrarater-Test retest	51°	3.7°	7.20%
		Intrarater-Test retest	Lumbar extension		
Bauer et al., (2016)	Inertial measurement unit (IMU)-systems	Intrarater-Test retest	20°	6.5°	32.50%
		Intrarater-Test retest	Lumbar lateral flexion		
Bauer et al., (2016)	Inertial measurement unit (IMU)-systems	Intrarater-Test retest	18°	4.6°	25.50%
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	17.3°	5.5°	31.70%
			Hip flexion		
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	54.7°	13.2°	24.10%
			Hip internal rotation		
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	2.54°	6.8°	267.70%
			Knee Valgus		
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	7°	2.7°	38.50%
			Knee flexion		
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	53°	10.2°	18.80%
			Knee internal rotation		
Alenezi et al. (2016)	Pro-Reflex, Qualisys	Within- & between-day reliability	5.2°	7.9°	151.90%

* UL, Upper Lumbar Spine; LL, Lower Lumbar Spine; MDC, Minimal Detectable Change

The third objective of the present study was to examine the reliability of the selected system which provides a novel method for measuring the multi-spinal regions and head-cervical region. It was hypothesised that, there will be a correlation between three scores of multi-regional spine range of motion, measured by a single examiner (within day reliability) using 3A sensors system. When the ICC of three measurements reaches > 0.80, the null hypothesis will be rejected. The Intra-class correlation coefficient (intra-tester reliability) for all regions was found to be high, ranging from a mean score of .88 (95% CI .62-.93) at the middle thoracic region during left rotation (Table 4.2.3) and .99 (95% CI .99-.99) at head-cervical region during left rotation (Table 4.2.2). Standard error for the measurements and MDC were between an SEM=0.4° and MDC=1.1° and an SEM=5.2° and MDC=14° for all movements and regions of the spine. A number of previous studies reported greater MDCs than produced during the present study (Table 5.3.1, 5.3.2 & 5.3.3). Kobler et al., (2013) and Bauer et al., (2016) concluded that their systems enable clinicians and researchers to objectively quantify the lumbar movement and movement dysfunction associated with LBP treatment efficiency. Accordingly, the 3A sensors can be claimed to be a valuable and portable system for measuring multi-spinal regions, however, this system can be used only for 'within day' measurement, either for assessment or measurement of difference between pre and post treatment sessions.

The study's findings are only representative of a young to middle age, healthy, 'male' population and reproduction with greater and more wide populations would be demanded to study the relationships in wider age groups and in females. In light of this, any discussions relating to the investigation and management of LBP populations, both gender population and other experimental study design other than intra-examiner 'within day' warrant careful consideration. This method is currently limited by the atypical method required to measure rotation, due to the need to align this plane with gravity, the errors due to skin movement under the sensors and experimental design (intra-examiner within day).

The second aim of this thesis was to quantify the relative contribution of five regions from within the thoracolumbar region as well as head-cervical region during flexion, extension, lateral flexion to right and to left, and rotation to right and to left in order to confirm the reliability of the selected system. This study has used a novel measurement method, which is unique in enabling a regional breakdown of the range of motion within a typical clinical setting. Measuring the regional breakdown of spinal motion in three planes and describing the relative motion of different regions of the thoracolumbar (TL) spine can provide useful clinical information, which can be used during routine clinical procedures for spinal assessment. Such information may allow clinicians to identify the regions of relatively altered mobility, to identify the appropriate plan and treatment. The relative contribution of number of spinal regions, which gathered from a multi-regional breakdown, adds insight into the relative contributions to spinal movement. Data acquisition, describing multiple spinal regions, enables the observation of the relative contribution of each region to the overall motion; thus, clinicians can now access a wealth of information regarding spinal movement behaviour. For example, the movement of extension displays up to four times greater movement at the lower spine, compared to the upper spine. The majority of extension occurs in the mid-thoracic region of the thoracic spine, with smaller contributions occurring from above and below. Subsequently, this allows the regions of relatively altered mobility to be identified and targeted for treatment, as changes in the relative contributions to motion are likely to alter the movement and loading behaviour of specific anatomical structures (Adams et al. 1980). The contribution information from multi-regional breakdown adds insight to the relative contributions to spinal movement behaviour, which was not previously accessible.

5.4 Contribution of the upper and lower lumbar spine, relative to hip motion, in dominant daily sagittal tasks

To investigate whether dividing the lumbar spine into two separate regions, relative to the hip region, yields a different understanding of the movement behaviour, compared with a traditional single joint, a series of experiments were undertaken. Single joint motion was measured during dominant functional tasks in the sagittal plane, which was then compared with a sectioned approach, where the lumbar spine was divided into two distinct regions, namely the upper lumbar and lower lumbar, during common daily functional activities. The results support the finding that the movement behaviour of the WL differs to that of the two smaller regions; therefore, greater understanding of lumbar movement behaviour can be gained from more complex sectioning of the lumbar spine into the upper and lower lumbar regions. As the whole lumbar spine consists of six spinal joints, the region between T12 and L3 was divided as the upper lumbar spine, while the region between L3 and S1 was divided as the lower lumbar spine. Traditionally, the lumbar spine has been divided as the region between T12 and S1. These three regions have been assessed per segment; the whole lumbar has six segments (vertebral joints), with the upper lumbar spine having three segments with the lower lumbar spine also having three (i.e. the WL/6 and UL/3 and LL/3). Dividing the lumbar spine into two distinct regions demonstrates that the per-segment range of motion was different, compared to the lumbar spine as a whole. This study was the first to conduct a method dividing lumbar segments to allow comparison between the whole lumbar and smaller regions and a comparison with hip during flexion, extension, object lifting, stand-to-sit and sit-to-stand. This method precludes direct comparison with previous studies; therefore measuring the lumbar spine as the UL and LL regions reveals that the 'per segment' contribution is different, compared to the lumbar region as one region, while contribution per segment of the LL was greater than the UL and WL over all tasks.

The average of contribution per segment of LL, UL and whole lumbar (WL) spine across all tasks were 10.4º, 6.5º and 8.4º, respectively. Therefore, the contribution per segment of LL was greater, on average, than WL, while WL was greater than UL over all tasks.

One task, such as flexion, the actual range of motion per segment of LL, UL and whole lumbar (WL) spine were 12º, 7.7º and 9.8º, respectively. The contribution per segment of LL was greater, on average, than UL by 4.3º, while it was greater than WL by 2.2º over all tasks. However, there is no standard value for such detailed measurements has been obtained up to that time, which mean this study is unique when it added new information of kinematics per segment. So far, there is no definite value for acceptable errors in vertebral joints even at lumbar spine, thoracic or cervical spine. In general, there is no standard level of error, which could be compared with the present study in terms of vertebral movement. It has been stated previously that the error of joint motion is prone to ±5° (Burton et al., 2000); however, this margin of error is related to a large range of motion joints, such as hip and knee joints, which cannot be accepted at a small joint such as the intervertebral joint.

Relying on validity findings, the axis error of 3A sensors, during sagittal plane motion of 360º was 3.63º. Accordingly, errors, which could be present at UL and LL spine movement in sagittal plane (i.e. when UL move 23º and LL move 30º during flexion) are about 0.15º and 0.2º, respectively. Such a magnitude of error is very small and it is anticipated that clinical researchers will adopt the measurement protocols to demonstrate the difference between per segment values of UL and LL spine. For instance, if there is no significant difference between 'per segment' motion of UL and per segment of LL spine during the flexion task, then this is an indicator of an injury, which occurs mainly at the LL spine. When measuring the difference between pre and post-intervention 'within day' sessions of treatment, and the examiner has not found any significant difference, that is an indicator of no improvement and vice versa. From a clinical perspective, these values demonstrate that this measure is sensitive to the expected changes as a result of clinical interventions, such as manipulation. Physiotherapists can, therefore, be confident that change, following intervention, is due to a range of motion variation, rather than sensor movement or measurement error. Such information will inspire physiotherapists to adopt multi-spinal measurements during assessment protocols to avoid under or overestimation, which take place when one measures one region of spine or measure the whole spine as a single region.

Range of motion percentages per segment of the LL showed more than that of UL over all five tasks (Figure 4.3.3). It was more evident that dividing the lumbar spine as a whole underestimated the motion of the lower lumbar and overestimated the motion of the upper lumbar spine. This underestimation, for the lower lumbar spine, may be as great as 37% and overestimation for the upper lumbar spine as much as 45%.

This finding is in agreement with previous studies, which have found that a regional breakdown of the lumbar spine yields a more detailed understanding of the relative contribution of each spinal region (Williams et al., 2010; Larding et al., 2011; Williams et al., 2012; Parkinson et al., 2013; Williams et al., 2014). The current study is the first to adopt a method of normalization, to enable a comparison between the WL and smaller regions, which precludes direct comparison of values found in the literature. The findings are similar, however, only to those studies adopting stereo-radiography (Pearcy et al., 1985) or cadaveric testing (Yamamoto et al., 1989), thus, contributing to the increasing body of evidence that suggests a non-uniform breakdown of range of motion contribution for the lumbar regions. Subsequently, this indicates that simply dividing the lumbar spine as a whole region may omit some important kinematic information and underestimate the contribution from the LL.

This study used a novel method to investigate the ratio of normalised lumbar motion relative to the hip. The results demonstrate some differences between the WL and the segmented lumbar spine with respect to ratios, suggesting that either region may be effective in exploring lumbar-hip ratios. Previous studies have explored lumbar-hip ratios, using a similar WL region, demonstrating ratio values similar to the current study for flexion (Lee and Wong, 2002; Wong and Lee, 2004; Shum et al., 2005) and values slightly greater than Shum et al. (2005) for sit-to-stand and stand-to-sit. This study demonstrated greater value for the ratio during extension of lumbar-hip (the mean values of WL was greater than the mean values of hip during extension), which are contrary to the findings in Lee and Wong (2002) and Wong and Lee (2004), both of which demonstrated greater hip contribution and therefore, smaller ratio. This may be due to differences in patient characteristics, such as the current study having exclusively investigated males, or due to a lower mean age, resulting in greater lumbar flexibility as displayed by the differences in lumbar extension range of motion (Lee and Wong, 2002; Wong and Lee, 2004).

Despite the lack of difference between the two spinal regions, there were differences between the UL and LL, suggesting the relationships between the hip and these specific lumbar regions are functionally different and unique. LL-hip ratios were consistently higher than the UL-hip ratios, due to the greater range of motion as demonstrated by the LL. This suggests that the relationship between the separate regions of the lumbar spine and hip were not equivocal and should be explored individually to appreciate the differences in kinematic behaviour.

The calculation of ratios in this manner provides insight only with regard to the relationship of the terminal ranges (i.e. the lumbar spine region and hip may move at the same speed at the initial phase and possibly lead one or the other to be delayed in the middle phase). Angle-angle plots can provide a description of where the range of motion of each region is plotted against one another, thereby, revealing further insights into kinematic behaviour.

Figure (4.3.4) illustrates the WL plotted against the hip and the UL-hip and LL-hip plots for comparison (the green line represents a 1:1 ratio for comparison). If a WL region was used, the behaviour would demonstrate that the hip and WL move at a similar time and rate throughout the movement phase (i.e. broadly correlating with the aforementioned green line). However, the regional breakdown shows a significantly greater contribution from the hip relative to the LL spine and such behaviour would not be visible with a WL region.

The findings from the current study suggest that regional breakdown of the lumbar spine is also important regarding velocity. Differences between the WL and regional spinal regions were detected, as were differences between the LL and UL. This suggests that the UL and LL are also functionally different regarding the higher order kinematics. The velocities determined in this study were slightly greater than those reported in other studies for movements at natural speeds (Shum et al., 2005; Williams et al., 2013). These differences may be due to differences in characteristics of the sample, such as age and sex (young-middle age males population) or the presence of pain (Williams et al., 2013). The additional information gained from the regional breakdown of the lumbar spine identified that the LL consistently moved at greater velocities. This is unique and important, since it suggests a difference in vertebral velocity between the upper and lower lumbar spine, a finding not discovered by a traditional single region analysis.

The findings of the current study have important clinical ramifications. Clinicians are beginning to advocate the assessment of two separate functional regions within the lumbar spine (O'Sullivan, 2005; Dankaerts et al., 2006), with the belief that these are functionally distinct. This study confirms that, indeed, there are functional differences in the range of motion of lumbar spine regions, and the relative velocities of motion, during a range of functional tasks, which provides support for the use of a more detailed spinal kinematic region. Greater contributions to motion from the lower lumbar spine, as well as greater movement velocities, may help to explain the increased prevalence of lower back pain or pathological change in this spinal region more than in the upper lumbar (BieringSørensen, 1983; Beattie et al., 2000). Usually, greater degeneration takes place in the lower lumbar spinal segments (Twomey and Taylor, 1987; Quack et al., 2007) and it is assumed that this is due to greater mechanical stresses being exerted upon this region as a result of greater mass effects (Adams and Hutton, 1983). Assessment of the lumbo-pelvic rhythm has also been suggested during clinical assessment of the back (O'Sullivan, 2005), as the hip motion affects the resultant bending stresses (Dolan and Adams, 1993) and muscle activities, as well as the forces acting on the lumbar spine (McGill et al., 2000; O'Sullivan et al., 2002; Kami.ska et al., 2010). Insights into lumbo-pelvic rhythm can be gained through the determination of ratios and angle-angle plots, and this study provides novel detail regarding the regional spinal ratios.

This study provides further evidence for the separation of the whole lumbar spine into smaller regional sections, as suggested previously (Parkinson et al., 2013), to truly determine detailed kinematic information for the lumbar spine.

To summarise the contribution of the upper and lower lumbar spine, relative to hip motion, this study was aimed to investigate whether dividing the lumbar spine into two separate regions, and analysing their motion relative to the hip will yield a different understanding of movement behavior, compared with a traditional single joint region during the dominant functional tasks in the sagittal plane. This aim has been achieved, due to implementing two objectives, to measure the range of motion and relative velocity magnitude using a traditional region of the lumbar spine as one single joint and then compare this approach with an analysis of smaller regions. The lumbar spine divided into two distinct regions, namely the upper lumbar and lower lumbar spine. These lumbar regions findings were compared with the relative motion of the hip. The findings showed differences between the UL and LL, suggesting the relationships between the hip and these specific lumbar regions are functionally different and unique. LL-hip ratios were consistently higher than the UL-hip ratios, due to the greater range of motion as demonstrated by the LL. This suggests that the relationship between the separate regions of the lumbar spine and hip were not equivocal and should be explored individually to appreciate the differences in kinematic behaviour. This study confirms that, indeed, there are functional differences in the range of motion of lumbar spine regions, and velocity of motion during a range of functional tasks, which provides support for the use of a more detailed spinal kinematic region. Greater contributions to motion from the lower lumbar spine, as well as greater movement velocities, may help to explain the increased prevalence of lower back pain or pathological change in this spinal region more than in the upper lumbar (BieringSørensen, 1983; Beattie et al., 2000). Known such that clinicians can identify the proper approach for lumbar problem. It is anticipated that this improved understanding of lumbar spine biomechanics may assist in developing a more effective treatment approach; for example, when a patient presents with a lumbar hyper-mobility, treatment should minimise intervertebral motion by providing stabilisation exercises, however, hypo-mobility of lumbar spine, would require mobilisation or any treatment which increases the ROM.

5.5 The correlation of lumbar-hip kinematics and flexion and other functional tasks

The aim of this study was to explore the relationships between the different sagittal tasks commonly assessed within the clinical environment, to determine if the resultant kinematics represent distinctly different movements. This study attempts to provide a more detailed investigation regarding the relationships between the flexion movement as a commonly assessed task within the clinical environment and different dominant functional tasks in the sagittal plane. This was achieved using a novel sensor string, enabling multiple anatomical regions to be studied. The ROM data were differentiated to yield velocity using Matlab codes to run a five-point differentiation to yield angular velocity. These codes were written and used by Williams et al., (2013) for angular velocity measurement.

The results of this study show that, the sagittal kinematics of the hip and lumbar spine during flexion are different from those observed during other functional tasks (i.e. lifting, stand-to-sit and sit-to-stand). This finding suggests that the movement of flexion is unique, compared to the other movements investigated. It is common for clinicians to assess flexion in a routine clinical examination of the spine; however, these findings imply that it may be necessary to assess other functional tasks in a more robust manner than is currently undertaken.

The results of the study show that there are similarities between flexion and lifting. At both lumbar regions, there was little difference in the range of motion used; indeed the magnitude of difference was less than 2º. However, movement variance appeared at the hip and was demonstrated to be 10º greater during lifting. This did not, however, appear to influence lumbar spine kinematics, suggesting that subjects who use more hip flexion during lifting do not necessarily decrease their lumbar contribution. However, participants in the present study appear not to routinely alter their lumbar curvature during low load lifting (3kg). This suggests that participants used spinal flexion during lifting, such as forward bending. In addition, individuals seem not to routinely alter their lumbar curvature during low load lifting, a finding observed previously within the literature (Williams et al. 2012; Williams et al. 2013; Parkinson et al. 2013). The contribution of three anatomical regions i.e. UL, LL and hip during stand-to-sit and sit-to-stand, are very similar with variance at flexion and lifting. Hip contribution at stand-to-sit and sit-to-stand appeared greater than that of other regions, which was also the case for flexion and lifting.

When regional angular velocity was been obtained using the aforementioned Matlab programme (Section 3.1.6), series of Matlab codes (Appendix C) ,) it was shown to be comparable to that available in the literature (Marras and Granata, 1997; Marras et al., 2000; Marras et al., 2001; Pal et al., 2007; Esloa et al., 1996; McClure et al., 1997; Shum et al., 2007; Williams et al., 2013). Lumbar velocity, during object lifting was found to be close to the findings of Marras and Granata (1997), Marras et al. (2000) and Marras et al. (2001). In addition, velocity during lumbar flexion was similar to that conducted by Consmuller et al. (2012) and slightly greater than that reported by Pal et al. (2007), Esloa et al. (1996) and McClure et al. (1997). The lumbar angular velocity, during lifting was found to be greater than that obtained by Shum et al. (2007) and Williams et al. (2013) who demonstrated their findings on LBP sufferers. These differences may be due to greater intra-participant movement variation, possibly due to having or expecting movement that would provoke pain. The differences that do exist could result from the different characteristics in the participants. Moreover, differences between findings in the literature may be due to number of studies have collected the data from LBP suffers, while other have collected the data from healthy participants. It is also likely that, due to age-related changes in the spine, the relationship between cardinal movements and functional movements are altered. The participants' ages or health conditions, could be the most important reasons of findings variation at previous studies. While it is known that the spinal motion and flexibility are associated with age, thus, the spinal movement and flexibility in younger people are more than those in advanced ages. The reason behind agreement between this study findings (i.e. movement and velocity) and (Marras and Granata 1997; Marras et al. 2000; Consmuller et al. 2012), could due to similar age and health condition. Participants in this study were asked to complete forward bending, backward bending, lifting an object (wooden box with handles weighing 3 kg) from the floor and to return to a standing position, moving from stand to sit on a stool and then returning to standing with no further instructions on how to move were provided. Without specific command during these tasks it is speculated that this could have produced different values, because every participant has decided to achieve the task using a preferable or an easiest way. They have achieved the task using different velocities (i.e. high, moderate or slow), particularly during return from lifting object from floor to upright standing. Although, the contributions of inter-participants show differences, the velocity average of this study was found to be close to the findings of Marras and Granata (1997), Marras et al. (2000) and Consmuller et al. (2012).

Thus, it can be suggested that the 3A system is capable of efficiently measuring angular velocity compared to the results of previous studies (Table 5.6.1).

Previous research has already provided substantial data about how manual therapy might increase spinal ROM (Powers et al., 2008; Goodsell et al., 2000; Lee et al., 2005). However, it has been proved that the ROM is not fully associated with function in LBP sufferers (Parks et al., 2003) and that LBP sufferers have great discrepancies in angular-velocities (Marras and Wongsam, 1986; Marras et al., 1995, 1999; Novy et al., 1999; Shum et al., 2007). Moreover, practitioners are increasingly interested in movement behaviours (Shum et al., 2005a; Dankaerts et al., 2007). This method of quantifying movement behaviour will enhance the understanding of movement at the lumbar spine, as well as enabling physiotherapists to study, in detail, the effect of interventions targeting movement control (Williams et al., 2013). The results of the current study demonstrate that different quantification of angular velocity is achievable for multi-spinal of lumbar spine relative to hip velocity using spatial plots.

Table 5.6.1. Comparison of velocity values at the lumbar spine from the literature

Study Methods	**Angular velocity (os-1)**	**Study Methods**	**Angular velocity (os-1)**
Marras and Granata, (1997)		Shum et al., (2007a)	
Lumbar motion monitor Lifting	47	Electromagnetic during picking up activity	
Healthy subjects		Healthy subjects	
Marras etal., (2000a)		picking up activity	30
Lumbar motion monitor lifting	48	LBP group	
Healthy subjects		picking up activity	19
Marras et al., (2001)		Williams et al., (2013)	
LMM during lifting		Inertial Sensors	
LBP group	21.3	Acute LBP	
Healthy subjects	36.5	Flexion	20
Pal et al., (2007)		Lifting	22
Opto-electronic during flexion		Extension	10
Healthy subjects		Chronic LBP	
Flexion	44	Flexion	28
Esloa et al., (1996)		Lifting	34
Opto-electronic during flexion phase		Extension	15
LBP group		Consmuller et al., (2012)	
Flexion	36	Epionics SPINE	
Healthy subjects		Healthy subjects	
Flexion	42	Flexion	54
McClure et al., (1997)		Extension	23
Opto-electronic during return from flexion phase		Present study	
LBP group		3A system	
Flexion	35	Healthy subjects	
Healthy subjects		Flexion	51
Flexion	31	Extension	32
		Lifting	56

Velocity demonstrated some distinct differences between the two movements for the LL and hip regions. Therefore, despite the range of motion being similar, suggesting similar kinematic profiles, it is in the higher order kinematics (velocity) that differences exist, which demonstrates that lifting resulted in greater velocity at the LL spine and hip. Whilst this finding has been reported previously, it suggests that providing an individual with a target or focus to the motion seems to result in greater velocity, consistent with the findings reported in (Williams et al. 2013). This is the first time such movement profiles have been measured in healthy subjects. It is clear that this new information permits the discovery of which part of the movement is affected. The ability of the spatial plots to display movement behaviour is of potentially great value to physiotherapists. The 3A sensors are easy to use and help to present information that can be analysed about the segmental movement and its associated velocity during clinical evaluation.

Correlation, as opposed to testing for differences, explores the relationship between the ROM across the tasks (rather than the difference in range of motion for each task) and the results suggest that only a moderate relationship in the range of motion was evident. A strong correlation between flexion and lifting was noted for the LL spine, suggesting a good relationship between the magnitudes of motion demonstrated between these two motions. This provides further evidence of the similarity in behaviour between these motions for the LL region. It is not known whether an alteration in one of these movement profiles will directly affect the other and whether this might be something worthy of further investigation. Only moderate correlations were noted for the UL and hip regions, providing evidence of a weaker relationship and illustrating a lack of similarity between these tasks for these regions.

Stand-to-sit and sit-to-stand appear to utilise different kinematic profiles for all anatomical regions. Compared to flexion, less spinal range of motion is evident with a greater contribution provided by the hips. These findings are supported by previous studies on both lumbar flexion and sit-to-stand, and stand-to-sit (Shum et al., 2005a). Furthermore, this study found a greater contribution from the LL spine during both flexion and sit-to-stand, and stand-to-sit. Previous studies have explored the relative motion between the lumbar regions during sit-to-stand only (Leardini et al., 2011; Parkinson et al., 2013); therefore, this study has expanded the analysis to include other functional tasks on lumbar regions and hip.

The inclusion of these functional tasks, during clinical assessment, will explore the different relationships between the lumbar spine and hip and is likely to provide different information about overall movement behaviour of the lumbar-hip region than flexion alone. Self-selected velocity for flexion compared with sit-to-stand and stand-to-sit provides further evidence of the uniqueness of these tasks. Flexion was consistently completed using greater velocity for the spinal regions, compared to sit-to-stand and stand-to-sit, with the opposite being true for the hip. Velocity during flexion seems to poorly correlate with velocity, utilised during other functional tasks, suggesting that each task has distinct properties relating to dynamic movement behaviour. The correlations between velocity of different tasks for the lumbar spine and hip have not previously been explored in the literature; therefore, this novel finding provides new insights into the relationship between flexion and other tasks. Velocity has been shown to be a key determinant of movement smoothness and therefore, provides important information regarding kinematics (Williams et al., 2013). Therefore, clinically, the inter-relationship between hip movement velocity and lumbar velocity cannot be fully explored using flexion alone.

This study suggests that the motion of flexion is unique in its kinematic profile, which puts forth the suggestion that clinicians should not be overly reliant on the interpretation of flexion range of motion within clinical environments to determine a degree of impairment. The results suggest that other sagittal tasks are unique in how they challenge the lumbar spine and hip and therefore, clinicians should be cautious about inferences drawn from assessing flexion alone. The failure to assess other functional movements relevant to the patient is likely to result in an incomplete understanding of the movement profile. An assessment incorporating other functional tasks, even if they are in the same movement plane, may be necessary to better understand the movement behaviour of these regions.

Strong correlations were only evident for the lower lumbar spine range of motion, between lifting and flexion, while all other tasks revealed moderate or weak correlations. Significant differences were evident in the range of motion and velocity used, when comparing flexion to other sagittal tasks.

On a cautionary note it must be stressed that clinicians not simply extrapolate the findings from clinical testing of flexion to other functional tasks since they demonstrate functionally unique kinematics.

To summarie the correlation of lumbar-hip kinematics between flexion and other functional tasks, the aim of the study was to explore the relationships between the different sagittal tasks commonly assessed within the clinical environment, to determine if the resultant kinematics represent distinctly different movements. The present study attempted to provide a more detailed investigation regarding the relationships between the flexion movement, as a commonly assessed task, within the clinical environment and different dominant functional tasks in the sagittal plane.

This study suggests that the motion of flexion is unique in its kinematic profile, which and that clinicians not be excessively reliant on the interpretation of flexion range of motion within clinical environments to determine a degree of impairment. The results suggest that other sagittal tasks are unique in how they challenge the lumbar spine and hip and therefore, clinicians should be cautious about making inferences from assessing flexion alone. The failure to assess other functional movements relevant to the patient is likely to result in an incomplete understanding of the movement profile. An assessment incorporating other functional tasks, even if they are in the same movement plane, may be necessary to better understand the movement behaviour of these regions.

Strong correlations were only evident for the lower lumbar spine range of motion, between lifting and flexion, while all other tasks revealed moderate or weak correlations. Significant differences were evident in the range of motion and velocity used, when comparing flexion to other sagittal tasks. Therefore, clinicians should not extrapolate findings from clinical testing of flexion to other functional tasks as they demonstrate functionally unique kinematics. The null hypothesis of this study was rejected as there were significant difference between flexion task and stand-to-sit task, as well as between flexion task and sit-to stand.

5.6 The limitations

5.6.1 Technological limitation:

The 3A sensor system was capable of measuring only two dimensions relative to gravitational inclination. This limitation restricted the rotational measurement during typical physiological positioning, therefore an atypical method was required to measure rotation, due to the need to align this plane with gravity. This has the potential to increase possibility of measurement errors. Even though this system is capable of measuring the spinal movement from a supine position to full rotational movement, this position was not used due the sensors potentially contact with bed surface leading to measurement errors. Whilst the SEM and MDC for rotational measurements was relatively high, viewed in the context of existing literature, it indicates that this error is consistent with other techniques and thus, reliability is greater in the two other planes. Whilst it is appreciated that the rotational methodology may be cumbersome or impractical for assessing some patients, the author remains confident that this represents a viable and convenient method for multi- regional spinal assessment for many patients within clinical practice.

5.6.2 Sample limitations:

It has to be acknowledged that the data set was exclusively collected from males; however, gender differences in thoracic shape and kinematic have previously been studied and the extent of these differences is typically small (Willems et al. 1996; Straker et al. 2008). This present study aimed to collect the data in the Cardiff School of Engineering population (males and females), however, participants who agreed to take part in this study were male only. To recruit volunteers for this research, Research Office Staff at Cardiff School of Engineering circulated an invitation email (including consent and research information sheets) several times to all staff members, researchers and students. Addition to invitation by email, researcher used to invite similar population orally. Furthermore, everybody has attachment of information sheet and consent form and handed over those forms manually during verbal invitation. However, no females responded to this invitation.

It is likely that, due to age-related changes in the spine, the relationship between cardinal movements and functional movements are altered, however, this study was limited by the sample represented. The young to middle age volunteers cannot be assumed to represent extremes of age. The results may not extrapolate to those outside the age ranges studied. A further limitation was that the participants were healthy, age ranged from 20 to 43 years and with weight values below 76.6 kg.

The experimental population were also, healthy and, therefore, serve as a reference for an asymptomatic population (Krawczky et al., 2014). Therefore, the results limited to an asymptomatic population and could be replicated in symptomatic population. However, it can be used these results produced similar values to previous studies (i.e. when comparing the total of regional values at lumbar spine or at thoracic spine with previous studies which measure each spinal region as one segment).

5.6.3 Biological limitations:

The study measurement system included six sensors fixed to the skin overlying the six spinous processes, and were applied whilst standing, which due to skin movement cannot be assumed to precisely represent the location of the relevant spinous process at the extremes of movement. Skin location devices, including inertial measurement systems are inherently prone to error, due to movement artifacts (Ha et al. 2013). Small movement of skin over spinous processes, will produce information errors. While, there is criticism regarding these spinal measurement systems, which place sensors directly on the skin, applications should give extra caution to attaching the sensor correctly which can be time-consuming.

Even the SEM and MDC for rotational measurements was relatively high; the MT, produced SEM=5.2° and MDC=14° for rotational measurements. However, the thoracic region is consistent with other studies, which have previously been attributed to the task difficulty (Ford et al. 2007). In addition, inter individual biological and flexibility differences across the general population may also be attributable to producing such error (Hopkins 2000). Limitations could have produced the difference between subject's movement contributions, for instance flexion of HC (min=42 & max=82°), UT (min=-14 & max=22°), MT (min=-4 & max=10°), LT (min=2 & max=36°), UL (min=2 & max= 32°) and LL (min=30 & max=53°).

5.6.4 Study design limitations:

During the present study, the difference between participants' values could not have been a result of biological factors or skin movement or even system inaccuracy only, but also could have been due to there being no specific instruction to perform the tasks. For instance, during the spinal flexion task, to perform this movement, the participants completed the movement in the most comfortable way, however, it has been noted that some subjects achieved a minimal curvature at the lumbar and thoracic regions, while majority of movement occur at the hip joint. The present study was limited also, by assessing intra-examiner 'within day' reliability and this may potentially limit the applicability of our findings in clinical settings between observers during a day.

5.7 Clinical implications

Evidences from both validity and reliability studies of the 3A sensor system has confirmed its feasibility when conducting spinal measurement. Although, the MDC was found to be high at two regional movements, due to biological factors, which increased the possibility of errors at these regions, the 3A sensors are considered a viable option as its ICCs were high for all regional movements and MDCs found smaller or similar to other studies. These studies have reported that systems which used for their experimental methods were useful for clinical implications (Table 5.3.1, Table 5.3.2 & Table 5.3.3). This system provide similar values to those obtained in previous studies findings whilst demonstrating better intra-rater reliability and MDCs no worse than those acceptable studies conducted previously. In addition the system benefits from being small in size and sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics. A number of motion systems have proven reliability, however, they are typically constrained to use in a laboratory context e.g. optical tracking systems and electromagnetic, or inertial sensors, which has not more than two sensors to capture the movement and velocity throughout accelerations. Inclinometer systems such as the iPhone, Bubble inclinometer and CROM device could not measure multi-spinal regions, while 3A sensors having a six string sensor array has the capability to measure five regions at same time. This system can be use in clinical applications, such as measuring the ROM and functional assessments (i.e. screen session), pre and post operatively and for assessing the 'immediate effect' of treatment, such as manipulation or mobilisation. However, it cannot be used for monitoring the improvement for more than one day, since it still needs to have its reliability between days confirmed.

This thesis presents new normative data, describing the kinematics of multi-spinal regions. Therefore, physiotherapists are now capable of using these data as reference for similar protocols of assessment to identify abnormal movements. Such data is crucial for analysing the range of motion of multi-spinal regions providing the opportunity to expand our perception with respect to assessing the severity of spinal disorders. For instance, the development of ankylosing spondylitis and the surgical influence on multiple-level discectomy or laminectomy (Hsu et al., 2008).

Demonstrating the ratio of LL-hip and UL-hip added a new procedure for spinal assessment and new information for physiotherapist to use more detailed findings as a normative data during spinal regional assessment at physical therapy clinics. For instance, when the LL region has been affected, the ratio of LL-hip and UL-hip during flexion may produce no significant difference, due to a decrease in both the LL ROM and velocity. Such a finding that the UL and LL are functionally independent is important for clinical practice and the application of an appropriate treatment. For instance, if the ratio value of UL-hip or LL-hip is high, that is an indicator of intervertebral hypermobility which may be treated by stabilising exercises, while a small value is an indicator of intervertebral hypo-mobility, which may require a program that increases ROM. However, clinicians' understanding that the lower lumbar regions contribution is greater than upper lumbar spines is very useful and subject to confirmation can be potentially implemented in clinics to assess the regional movement and relative functional behavior.

The new information discovered that the sagittal kinematics of the hip and lumbar spine, during trunk flexion, are different from those observed during other dominant functional tasks in the same plane. This conclusion could change physiotherapy protocols for spinal assessment by adding more tasks, such as standing to sitting and sitting to standing and suggests that physiotherapists should not simply rely on flexion assessments alone.

Chapter 6: Conclusions

6.1 Conclusions

Spinal range of motion measurement is a routine clinical procedure, where normal or abnormal range of motion, movement behaviours (i.e. movement quality) and a base line are established for observing the immediate treatment effects and functional improvements. An understanding of the normal physiological movement of lumbar spine regions and hip, as well as the behaviour of each regional movement, during dominant daily tasks in sagittal plane, is an essential prerequisite to clinical diagnosis and treatment. Thus, the selection of an appropriate measurement system, which is capable of measuring dynamic movement in 'real time' was a fundamental aim of this study. This required assessment of both validity, against a "gold standard" system and reliability, by measuring the range of motion of multi-spinal regions.

The first objective which was established was to select an appropriate system, capable of capturing spinal kinematics. Applying a number of scientific criteria this was achieved when findings from an evaluation process for number of systems determined that the 3A sensor system recorded the highest value. The 3A sensors system, which gives orientation and acceleration information with gravitational orientation, was selected; based on being superior to other systems across a range of assessment criteria.

Other systems, both non-invasive and invasive, appeared to be generally impractical by comparison for both the average clinicians and even research laboratories, due to their respective limitations with respect to accuracy, portability, setting, calibration, accessibility, time-consumption, constrained field of view and cost. The 3A sensor based system was novel in the following ways:

- The use of accelerometry for spinal motion analysis is still in its infancy.
- Measuring regional effects was sparsely explored (regional lumbar, regional thoracic).
- The system was capable of a clinically meaningful application - no special processing required.
- The system constituted multiple linked (chained) sensors operating simultaneously, not just two or three sensors.

However, even though this system was capable of measuring the angulation movement, velocity and acceleration precisely, this technology is limited to measure two dimensions relative to gravity.

A valid and reliable clinical method for measuring spinal kinematics of many regions was designed to overcome the limitations of current systems. The validity and reliability of any measurement system should be proved, such that it can be used in clinical practice, research, or both. Therefore, two studies were conducted to confirm selection of the tri-axial accelerometer sensors based system. Validation was achieved against a "gold standard" rolly table, which revealed a high correlation between the two systems and an acceptable rate of error, compared with previous studies. The evidence from this study suggests that the sensor system is capable of measuring spinal movement, both in the clinical and research fields. This system is sufficiently cost-effective for multiple sensors to be used along the length of the spine to determine segmental spinal kinematics, small in terms of size, portable and capable of measuring movement in real-time whilst being relatively easy to use.

Moreover, the reliability of this system has also been confirmed, by measuring the multi-spinal regions range of motion and demonstrating excellent reliability and different values of MDCs, thereby providing a viable and practical method for assessing a number of multi-regional clinical spinal motions.

The relative motions of multi-spinal regions, during flexion, extension, lateral flexion to right and to left and rotation to right and to left has added insight and a newly-established understanding of spinal movement. Furthermore, the method provides valuable information, which can assist clinicians in assessing, which region is subjected to the greatest mechanical problems, based on the regional contribution. Such information will save clinicians' time and act as a guide in identifying problems and selecting suitable treatment approaches. Although, the MDCs were high at two regional movements due to biological difference among participants, which increased the possibility of errors at these regions, as well as skin movement over bones, the 3A sensors provide a viable option since its ICCs were found to be high for all the regional movements and the MDCs found to be smaller or similar in range to previous approaches. The system can be used in clinical applications, such as measuring ROM and functional assessments (i.e. screen session), pre and post operations and assessing the immediate effect of treatment, such as manipulation and mobilisation. Therefore, physiotherapists are now capable of using the data as reference for similar protocol of assessment to identify abnormal movements. This understanding is crucial, because analysing the range of motion of multi-spinal regions provides the opportunity to expand our perception of the relative severity of spinal disorders. For example, the development of ankylosing spondylitis and the surgical influence on multiple-level discectomy or laminectomy.

Obtaining kinematic information for the lumbar spine in more than one region (i.e. upper and lower lumbar spines), relative to hip kinematics during daily functional tasks, such as flexion, extension, lifting, stand-to-sit and sit-to-stand, could provides essential information regarding the kinematic interactions of the lumbar-hip complex.

Understanding the relationship between forward flexion (i.e. cardinal motion) and these tasks could produce two options: that the relationship between flexion and other tasks is strong, which means that evaluation of these tasks becomes unnecessary in clinical assessments and deriving weak correlations, between flexion and other tasks, which indicates a new trend in which spinal assessment protocols may change. This suggests that to measure the lumbar spine as a whole may risk missing out some important kinematic information. The findings of the current study suggest dividing the lumbar spine into two distinct regions to demonstrate normalised kinematic differences, rather than treating the lumbar spine as a whole.

It is evident that dividing the lumbar spine as a whole entity underestimates the contribution of the lower lumbar and over-estimates the contribution of the upper lumbar spine. Physiotherapists should be aware of the differences between the regions to better inform their clinical assessment of the lumbar spine.

The correlations and differences between the flexion and other dominant functional tasks indicates that the sagittal tasks utilise different lumbar-hip kinematics and place different demands on the spine and hip.

This study suggests that the motion of flexion is unique in its kinematic profile and suggests that physiotherapists should not be over reliant on the interpretation of flexion range of motion within the clinic to determine the degree of impairment.

The results suggest that other sagittal tasks are unique in how they challenge the lumbar spine and hip, and therefore, physiotherapists should be cautious about inferences made from assessing flexion alone. The failure to assess other movements, functionally relevant to the patient, is likely to result in an incomplete understanding of the movement profile potentially resulting in sub optimal treatment or even misdiagnosis.

An assessment incorporating other functional tasks, even if they are in the same movement plane, may be necessary to better understand the movement behaviour of these regions.

In general, these findings have provided a new, viable, valid and reliable clinical method for measuring spinal kinematics, including velocity, of many regions simultaneously, overcoming limitations of current systems; providing information that can be adopted and taken into account by clinicians when applying clinical practices. The sensor system was used to determine: a moderate correlation between flexion and lifting for spinal ROM, but not other sagittally dominant movements. No correlation for velocity was established, suggesting a different kinematic profile for sagittally dominant movements. Furthermore, the study has demonstrated the importance of a more regional approach to spinal kinematics. Such findings could contribute to the development and improvement of diagnostic services in medical clinics, such as physical therapy clinics. The study describes a large data set for spinal kinematics in normal males, this is important for regions not well explored, that is, the thoracic spine, but also lower lumbar and upper lumbar.

Chapter 7: References

7 References

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Appendix

Appendix A- list of publications

Paper-one

Reliability of an Accelerometer-Based System for Quantifying Multiregional Spinal Range of Motion

Raee S. Alqhtani, MSc,[a,b] Michael D. Jones, PhD,[c] Peter S. Theobald, PhD,[d] and Jonathan M. Williams, PhD[e]

Abstract

Objectives: The purpose of this study was to investigate the reliability of a novel motion analysis device for measuring the regional breakdown of spinal motion and describing the relative motion of different segments of the thoracolumbar (TL) spine.

Methods: Two protocols were applied to 18 healthy participants. In protocol 1, 2 sensors were placed on the forehead and T1 to measure cervical range of motion (ROM). In protocol 2, 6 sensors were placed on the spinous processes of T1, T4, T8, T12, L3, and S1 to measure TL regional ROM. Intraclass correlation coefficients were used to evaluate the repeatability of movement, whereas SEM was used to define the extent of error. Ranges of motion were demonstrated in flexion extension, right-left lateral flexion, and right-left rotation of the head-cervical, upper thoracic, middle thoracic, lower thoracic, upper lumbar, and lower lumbar.

Results: The intraclass correlation coefficient values, for all regions, were found to be high, ranging from 0.88 to 0.99 for all movements, and regions of the spine and SEM values ranged from 0.4° to 5.2°. Multiregional spine ROM ranged from 3° in the upper thoracic and mid-thoracic during flexion and 80° at head cervical during right rotation.

Conclusion: The described methodology was reliable for assessing regional spinal ROM across multiple spinal regions while providing the relative motions of different segments of the TL spine. (J Manipulative Physiol Ther 2015;38:275-281)

Key Indexing Terms: *Reliability; Spine; Regional; Range of Motion; Accelerometry*

Measurement of spinal range of motion (ROM) is common within the assessment of spinal disorders.[1] There are many methods to noninvasively measure spinal ROM, including simple clinical methods and more complex laboratory systems. The former includes techniques such as goniometry, inclinometry, and the cervical ROM device; however, all are only able to provide a single point in time, meaning movement behavior across time is lost.[2] Furthermore, measurement of 3 planes of motion is difficult and time consuming and typically demands complex laboratory methods. Optoelectronic methods are commonly used to measure ROM in 3 dimensions,[3,4] although they are time consuming, and data processing can be complex.[5] Electromagnetic systems have been used to measure spinal ROM in the cervical,[6] thoracic,[7] and lumbar spine,[8] although small operating fields and metallic disturbance in areas where metals are present should be considered as limitations.[9,10] Inertial sensors have quantified cervical[11] and lumbar ROM,[2] although the application of these methods to the spine often involves the use of 2 sensors, creating a hypothetical single "joint" of interest.[12-14] Here, the inherent limitation is that the distribution of movement across the length between the 2 sensors is unknown,[15] which is critical to understanding motion sharing within the spine. Furthermore, separate spinal regions are often studied in isolation, unlike in clinic where the aim is often to simultaneously assess multiple regions.

[a] PhD Student, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[b] Physiotherapy Specialist, Ministry of Health, Kingdom of Saudi Arabia.
[c] Senior Lecturer, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[d] Lecturer, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[e] Senior Lecturer, Faculty of Health and Social Sciences, Bournemouth University, Dorset, UK.
Submit requests for reprints to: Raee S. Alqhtani, MSc, 137 Admiral House, Newport Road, Cardiff, UK CF24 0DP.
(e-mail: AlqhtaniRS@cardiff.ac.uk).

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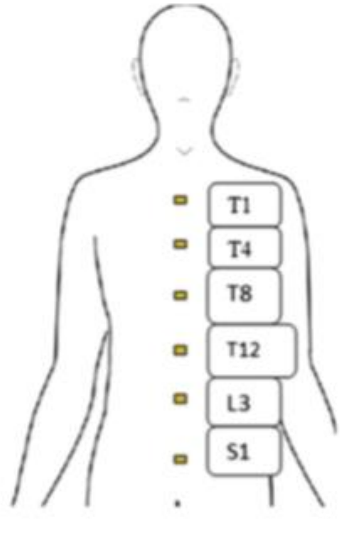

Fig. 2. *Schematic representing the location of spinal sensors.*

calculated. Matrix laboratory (R2013a; Matlab) was used to measure the relative motion during flexion extension, right-left lateral flexion, and axial rotation (right and left). Regional ROM (ie, HC, UT, MT, LT, UL, and LL) is equal to peak ROM between upper and lower sensor, relative to each region.

The movement of skin over spinous processes of vertebrae cannot be avoided when using external device such 3AS, and this error is likely to be systematic, hence leading to a relatively persistent bias in the obtained results.[20,21] However, the fascia over the spinous processes is quite firmly adhered to bone, which make the skin movement follow spinous processes motion more closely than in many other parts of the body.[22,23]

Intraclass correlation coefficients (ICCs) were calculated across the entire cohort using the Statistical Package for the Social Sciences software (Statistics 20), to evaluate the repeatability of the 3 repetitions recorded for each movement. The ICC value is recognized to provide a measure of repeatability,[24] with values classified using the following thresholds: less than 0.4, poor; 0.4 to 0.75, fair to good; more than 75, excellent as per the literature.[25] Confidence intervals (CIs), with a significance value of $P < .05$, were also calculated to provide a further statistical measure.

The standard error of measurement (SEM) was used to define the extent of error, meaning that greater reliability is defined by a smaller SEM value.[24] SEM was calculated using the formula: SEM = SD$\sqrt{}$ (1-ICC). SEMs presented in the Results section represent the range of values calculated for each variable, of each participant.

Results

The ICC values for all regions were found to be high (0.88-0.99, Table 1), demonstrating highly reliable data. This is further supported by considering the CIs, which describe broadly consistent data (Table 1). SEM values (Fig 3) for flexion/extension and right/left lateral again demonstrated a reliable measurement protocol (Table 2), although upper and mid-thoracic provided the highest SEM when measuring rotation (3.2° and 5.2°, respectively), meaning a potential relative error of approximately 20% when compared with the ROM data (Table 3). Multiregional spine ROMs ranged from 3.9° at MT during flexion, to 80° at HC during right rotation.

The regional breakdown of relative motion of the TL spine demonstrates that 47% of the flexion motion occurred at the LL and 41% for extension, representing the largest contribution from all the segments. Lateral bending relative motion demonstrates a more even spread of movement over the LL, UL, and LT spine with each region contributing 24% to 26% of motion. The MT region demonstrated the greatest contribution to rotation motion, with 36% to 40% for left and right rotation, respectively. A breakdown of relative contribution is displayed in Figure 4.

Discussion

This study primarily aimed to investigate the reliability of a novel motion analysis device for measuring the regional breakdown of spinal motion, before describing the relative motion of 5 TL segments. The results indicated that the novel device and protocol provide a reliable method for measuring multi-regional spinal ROM. This is evidenced by high ICC values (ie, >0.88) for repeatedly measuring each variable across the cohort. Indeed, the ICC values presented in this study compare favorably with other methods of spinal motion measurement including electromagnetic,[15,26-28] inertial,[11,29] and optoelectronic sensors.[30,31] This is the first study, however, that reports reliability values for specific spinal regions, indicating that reliability is not compromised when measuring smaller spinal regions.

The reported SEM were small, ranging from 0.4° to 5.2°, which typically represented a low relative error vs equivalent ROM data. Although there are no definite thresholds for acceptable SEM, it should be noted that acceptable goniometry data are typically susceptible to ±5° error,[24] meaning these data are no worse than that achievable in a single plane. From a clinical perspective, these values demonstrate that this measure is sensitive to the change expected from clinical interventions, such as manipulation. Clinicians can therefore be confident that change after intervention is due to ROM variation, as opposed to sensor movement or measurement error.

Measuring thoracic rotation caused the greatest SEM (ie, SEM >3°), which is consistent with other studies and has previously been attributed to the nature of movement (ie, task difficulty),[32] biological, and flexibility difference across the general population.[33] In this instance, it may,

Table 1. *Intraclass Correlation Coefficients With 95% CI for Multiregional Spinal Movement*

Spinal Regions	Flexion ICC (95% CI)	Extension ICC (95% CI)	Right Lateral Flexion ICC (95% CI)	Left Lateral Flexion ICC (95% CI)	Right Rotation ICC (95% CI)	Left Rotation ICC (95% CI)
HC	.98 (.95-.99)	.98 (.97-.99)	.98 (.95-.99)	.98 (.96-.99)	.97 (.93-.99)	.99 (.99-.99)
UT	.99 (.94-.99)	.96 (.91-.98)	.92 (.82-.97)	.98 (.95-.99)	.96 (.92-.98)	.99 (.98-.99)
MT	.97 (.94-.99)	.92 (.81-.97)	.91 (.90-.97)	.97 (.92-.99)	.99 (.97-.99)	.88 (.62-.93)
LT	.97 (.95-.99)	.92 (.83-.97)	.95 (.88-.98)	.95 (.88-.98)	.98 (.95-.99)	.99 (.98-.99)
UL	.98 (.95-.99)	.97 (.93-.99)	.97 (.92-.99)	.98 (.95-.99)	.97 (.93-.99)	.98 (.96-.99)
LL	.95 (.89-.98)	.96 (.91-.98)	.94 (.86-.98)	.90 (.78-.96)	.99 (.97-.99)	.98 (.96-.99)

CI, confidence interval; *HC*, cervical; *ICC*, intraclass correlation coefficient; *LL*, lower lumbar; *LT*, lower thoracic; *MT*, middle thoracic; *UL*, upper lumbar; *UT*, upper thoracic.

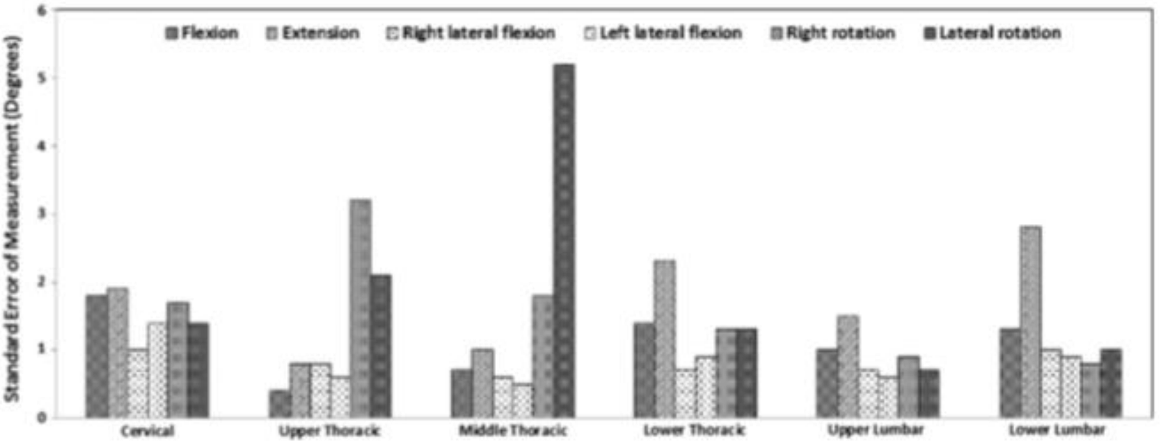

Fig. 3. *The SEM (degrees) for each spinal region during the 6 movements.*

Table 2. *SEM (Degrees) for Multiregional Spinal Movement*

Spinal Regions	Flexion	Extension	Right Lateral Flexion	Left Lateral Flexion	Right Rotation	Left Rotation
HC	1.8	1.9	1.0	1.4	1.7	1.4
UT	0.4	0.8	0.8	0.6	3.2	2.1
MT	0.7	1.0	0.6	0.5	1.8	5.2
LT	1.4	2.3	0.7	0.9	1.3	1.3
UL	1.0	1.5	0.7	0.6	0.9	0.7
LL	1.3	2.8	1.0	0.9	0.8	1.0

HC, cervical; *LL*, lower lumbar; *LT*, lower thoracic; *MT*, middle thoracic; *UL*, upper lumbar; *UT*, upper thoracic.

although, be due to the inherently greater movement variability at this spinal region or represent slight difficultly in the ability of the clinician to fix nonmoving regions as was necessary in this measurement of rotation.

This novel measurement method is unique in enabling a regional breakdown of ROM within a typical clinical setting. The actual ROM values were similar to those reported previously for regional thoracic spine,[7,34,35] total lumbar spine,[5,36] and also for the 2 smaller regions of the lumbar spine.[15,35] Data acquisition describing multiple spinal regions enables the observation of the relative contribution of each region to the overall motion; thus, clinicians can now access a wealth of information regarding spinal movement behavior. As an example, the movement of extension displays up to 4 times greater movement at the lower, compared with the upper, lumbar spine. Most extension occurs in the mid-thoracic region of the thoracic spine, with smaller contributions occurring from above and below. Subsequently, this allows the regions of relatively altered mobility to be identified and targeted for treatment, as changes in the relative contribution to motion are likely to alter the movement and loading behavior of specific anatomic structures.[37]

Limitations

This study was limited by the atypical method required to measure rotation, due to the need to align this plane with gravity. Although the SEM for rotational measurements was relatively high, viewed in the context of existing literature indicates that this error is consistent with other techniques, and, thus, reliability is greater in the 2 other planes. We appreciate that the rotational methodology may be cumbersome or impractical for assessing some patients; however, we remain confident that this may potentially be a method for multi-segmental spinal assessment for patients within a clinical environment.

Table 3. *Mean (SD) ROM (Degrees) for Multiregional Spinal Movement*

Spinal Regions	Flexion	Extension	Right Lateral Bending	Lift Lateral Bending	Right Rotation	Left Rotation
	ROM (SD)	ROM (SD)	ROM (SD)	ROM (SD)	ROM (SD)	ROM (SD)
HC	66.4 (13)	61.7 (11)	41.5 (7)	42.1 (10)	74.4 (10)	80.5 (14)
UT	3.9 (4)	7.1 (4)	6.5 (3)	5.4 (4)	−14.9 (16)	−11.3 (21)
MT	3.5 (4)	11.2 (8)	7.8 (2)	7.1 (3)	34.8 (18)	29.7 (18)
LT	15.0 (8)	7.9 (6)	12.1 (3)	12.4 (4)	21.4 (9)	22.6 (13)
UL	19.4 (7)	5.0 (9)	12.6 (4)	11.3 (4)	6.3 (5)	5.3 (5)
LL	36.8 (6)	21.6 (14)	12.2 (4)	11.6 (3)	9.4 (8)	8.7 (7)

HC, cervical; *LL*, lower lumbar; *LT*, lower thoracic; *MT*, middle thoracic; *ROM*, range of motion; *UL*, upper lumbar; *UT*, upper thoracic.

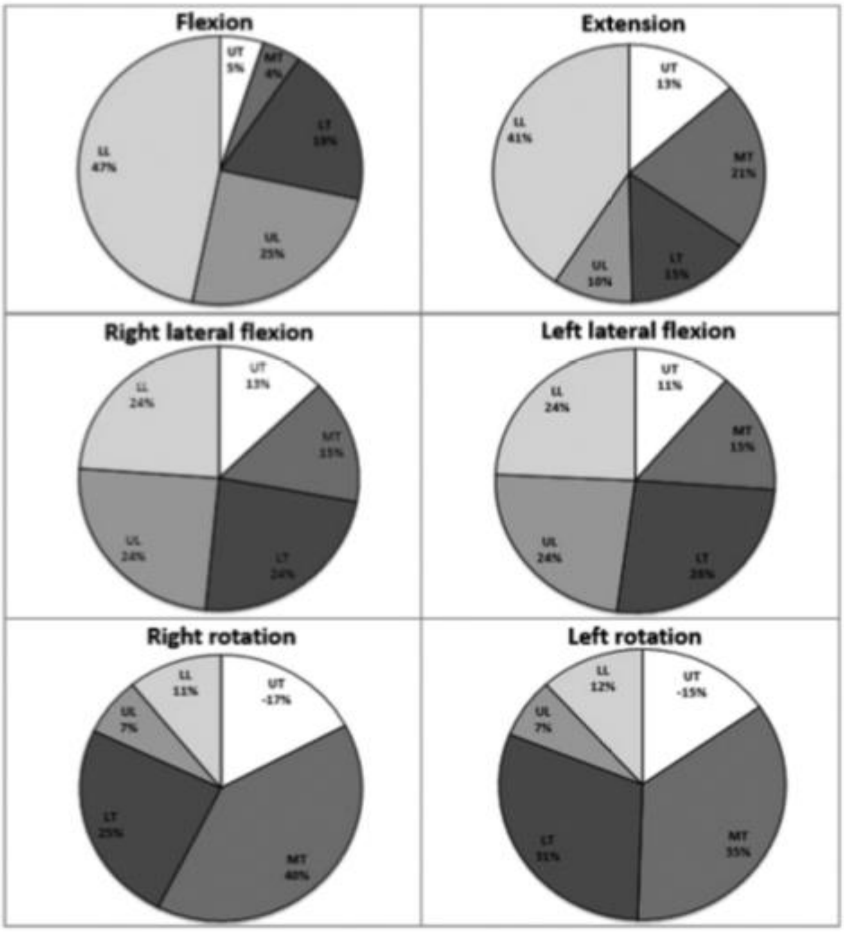

Fig. 4. *The percentage contribution from each spinal region during the 6 movements: HC, UT, MT, LT, UL, and LL. HC, cervical; LL, lower lumbar; LT, lower thoracic; MT, middle thoracic; UL, upper lumbar; UT, upper thoracic.*

Future studies could observe the interactions between spinal regions or investigate the reliability in clinical populations. However, we would encourage caution when interpreting rotational change in the upper and mid-thoracic region.

Conclusion

This multi-accelerometer system demonstrated excellent reliability and small errors to provide a viable and, largely practical, method of assessing multiregional clinical spinal motion. The additional movement information from multiregional breakdown adds insight into the relative contributions to spinal movement behavior, which was not previously accessible.

Funding Sources and Potential Conflicts of Interest

This research was funded by Saudi Arabia government. No conflicts of interest were reported for this study.

Contributorship Information

Concept development (provided idea for the research): R.S.A., M.D.J., P.S.T.
Design (planned the methods to generate the results): R.S.A., M.D.J., P.S.T.
Supervision (provided oversight, responsible for organization and implementation, writing of the manuscript): J.M.W., P.S.T., M.D.J.
Data collection/processing (responsible for experiments, patient management, organization, or reporting data): R.S.A.
Analysis/interpretation (responsible for statistical analysis, evaluation, and presentation of the results): R.S.A., M.D.J., J.M.W., P.S.T.
Literature search (performed the literature search): R.S.A., J.M.W.
Writing (responsible for writing a substantive part of the manuscript): R.S.A.
Critical review (revised manuscript for intellectual content, this does not relate to spelling and grammar checking): J.M.W., M.D.J., P.S.T.

Practical Applications

- The ICC values, for all regions, were found to be high, ranging from 0.88 and 0.99 for all movements, and regions of the spine and the SEM values ranged from 0.4° to 5.2°. Multiregional spine ROM ranged from 3° at UT and MT during flexion and 80° at HC during right rotation.
- The described methodology was shown to be reliable for assessing regional spinal ROM, across multiple spinal regions while providing the relative motions of different segments of the TL spine.

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Paper-two

Correlation of Lumbar-Hip Kinematics Between Trunk Flexion and Other Functional Tasks

Raee S. Alqhtani, MSc,[a,b] Michael D. Jones, PhD,[c] Peter S. Theobald, PhD,[d] and Jonathan M. Williams, PhD[e]

Abstract

Objective: The purpose of this study was to explore the relationship between the kinematic profiles of flexion of the upper lumbar and lower lumbar (LL) spine and hip and 3 sagittally dominant functional tasks (lifting, stand-to-sit, and sit-to-stand).
Methods: Fifty-three participants were recruited for this study. Four sensors were attached to the skin over the S1, L3, T12, and lateral thigh. Relative angles between adjacent sensors were used to quantify the motion for the hip, LL, and upper lumbar spine. Pearson correlation coefficients were used to explore the relationship between the movements and more functional tasks. One-way analysis of variance was used to determine the significance of differences between the variables.
Results: Flexion resulted in a greater or similar range of motion (ROM) to the other tasks investigated for both spinal regions but less ROM for the hip. Strong correlations for ROM are reported between forward flexion tasks and lifting for the LL spine ($r = 0.83$) and all regions during stand-to-sit and sit-to-stand ($r = 0.70$-0.73). No tasks were strongly correlated for velocity ($r = 0.03$-0.55).
Conclusion: Strong correlations were only evident for the LL spine ROM between lifting and flexion; all other tasks afforded moderate or weak correlations. This study suggests that sagittal tasks use different lumbar-hip kinematics and place different demands on the lumbar spine and hip. (J Manipulative Physiol Ther 2015;38:442-447)
Key Indexing Terms: *Flexion; Lifting; Sitting; Standing; Lumbar; Hip; Correlation; Function; Tasks*

Clinical evaluation of the lumbar-hip complex is commonplace in musculoskeletal therapies such as physical medicine/rehabilitation, chiropractic, osteopathic, and physiotherapy clinics.[1,2] Traditional texts advocate the assessment of motion in the cardinal planes. The evaluation of the behavior of the spine and hip during spinal motions such as flexion/extension is a potential test used to observe lumbar impairments.[3-5] Clinicians use the results of motion tests such as forward flexion to aid in the clinical reasoning process when attempting to determine treatment and rehabilitation options.

Disorders of the lumbar-hip complex have been shown to affect lumbar spine and hip range of motion (ROM) as well as the interaction between these 2 anatomical regions.[5-7] Moreover, disorders of the lumbar-hip complex have a demonstrably significant effect on movement velocity, both at the hip and at the lumbar spine.[8-13] This has been determined for cardinal ROM (lumbar flexion/extension) and in more functional movements such as lifting an object from the floor, a commonly reported daily activity.[11] Moreover, sit-to-stand and stand-to-sit are common activities of daily living, which are reportedly completed approximately 60 times a day in certain working populations.[14] These activities are also known to be affected by the presence of disorders of the lumbar-hip complex. This suggests that disorders of the lumbar-hip complex may affect functional tasks as well as the cardinal movements often used in the clinic.

[a] Student, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[b] Physiotherapy Specialist, Ministry of Health, Riyadh, Kingdom of Saudi Arabia.
[c] Senior Lecturer, Cardiff University, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[d] Lecturer, Cardiff University, Institute of Medical Engineering & Medical Physics, Cardiff School of Engineering, Cardiff University, Cardiff, UK.
[e] Senior Lecturer, Faculty of Health and Social Sciences, Bournemouth University, Dorset, UK.
Submit requests for reprints to: Raee S Alqhtani, MSc, Student, 137 Admiral House, Newport Road, Cardiff, UK, CF24 0DP. (e-mail: *AlqhtaniRS@cardiff.ac.uk*).
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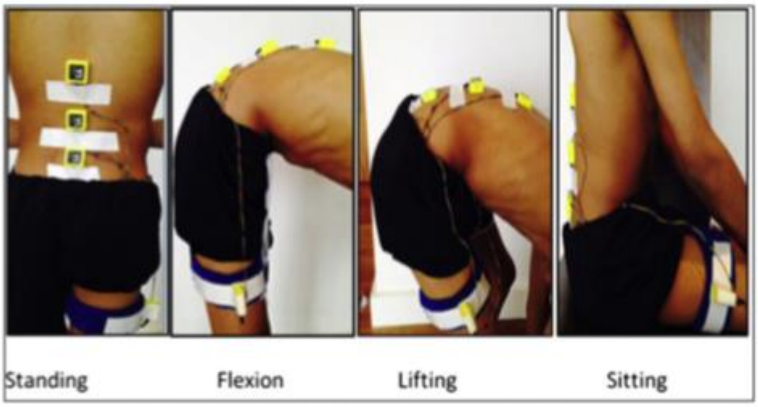

Fig 1. *Schematic represents the location of 4 sensors on spinous processes of T12, L3, and S1 and on the lateral aspect of the thigh midway between the lateral epicondyle and greater trochanter on the iliotibial band.*

Currently, it is not well understood to what degree the cardinal motions, such as forward flexion, are related to more functional tasks. It is possible that there is no relationship between forward flexion and other sagittally dominant functional tasks, such as lifting, stand-to-sit, or sit-to-stand. If there were no relationship, using forward flexion as a basis for exploring sagittal movement behavior would be flawed, potentially leading to erroneous clinical judgements and reasoning. However, it may be that forward flexion is closely related to other sagittal tasks, making the assessment of many tasks within the clinic unnecessary. Therefore, a better understanding of the relationship between forward flexion and sagittal tasks may aid in the interpretation of clinical assessment and treatment decision making.

The assessment of the spine usually involves the completion of movements in the cardinal planes, and the relationship between these cardinal motions and functional tasks such as lifting, stand-to-sit, and sit-to-stand has yet to be established. Therefore, the purpose of this study was to explore the relationship between the kinematic profiles of trunk flexion and 3 sagittally dominant functional tasks (lifting, stand-to-sit, and sit-to-stand). The kinematic profile for the anatomical regions of upper lumbar (UL) and lower lumbar (LL) spine and hip will be used to determine correlations and differences.

Methods

Subjects

Fifty-three subjects were recruited from Cardiff University (age, 29.4 ± 6.5 years; mass, 75.3 ± 16.4 kg; height, 1.69 ± 0.15 m). None of the participants had a history of spinal pain or reported any disorder of the cervical, thoracic, or lumbar spine or the hip. Participants were screened to be free from neurologic conditions, vestibular disturbances, inflammatory joint disease, and a history of spinal surgery. This study was approved by the Cardiff School of Engineering Ethics Committee. Participants were recruited via email advertisement to staff and postgraduate students; thus, our cohort was a convenience-based sample. All participants provided written informed consent.

Instrumentation

A string of 4 accelerometers (3A Sensors; THETAmetrix, Waterlooville, UK) was used to measure the kinematics of the lumbar spine and hip. Each sensor footprint was 24 mm^2 and was connected to a laptop computer via universal serial bus cable. Each sensor provides absolute orientation (tilt) with respect to gravity. Such a system has been shown previously to have excellent repeated-measures reliability relating to spinal motion analysis, with the intraclass correlation coefficient ranging from 0.88 to 0.99 and a standard error of measurement ranging from 0.4° to 5.2°.[15] The accuracy of such a system has been established in a preliminary study and shown to offer root mean square errors of 0.70% to 1.39% compared with a precision angle measurement table (THETAmetrix).

Procedure

Subjects were asked to perform a warm-up exercise, which included flexion, extension, and rotation of the trunk. Four sensors were placed firmly on the skin using double-sided hypoallergenic tape over the spinous processes of T12, L3, and S1 as well as the lateral aspect of the right thigh midway between the lateral epicondyle and greater trochanter on the iliotibial band (Fig 1). Participants were permitted 1 trial of the movements before data collection to familiarize themselves with the procedure and moving with the sensors attached. Participants stood barefoot on assigned markers and focused on a wall marker set at a height of 2 m with arms relaxed by their side. Movements included forward bending, lifting an

Table 1. *Mean (SD) ROM and Velocity for the 4 Tasks and Each Anatomical Region*

	ROM (degrees)			Positive Velocity (degrees/s)			Negative Velocity (degrees/s)		
Tasks	UL	LL	Hip	UL	LL	Hip	UL	LL	Hip
Flexion	23.3 (10.1)	36.0 (13.3)	53.2 (14.6)	22.4 (8.8)	31.6 (14.1)	33.0 (18.5)	22.2 (9.9)	28.7 (13.6)	35.0 (16.9)
Lifting	21.6 (9.9)	35.4 (13.9)	63.2 (14.6)	25.2 (11.8)	35.6 (13.4)	51.5 (22.4)	23.3 (8.8)	33.4 (14.5)	50.6 (25.3)
Stand-to-sit	17.0 (10.1)	27.0 (14.9)	64.4 (17.3)	16.6 (7.7)	26.7 (15.2)	57.5 (21.3)	10.0 (4.1)	16.3 (9.6)	35.0 (21.5)
Sit-to-stand	16.3 (10.2)	26.6 (14.9)	64.8 (18.4)	9.5 (5.8)	16.4 (10.6)	40.9 (22.2)	17.0 (8.6)	27.5 (15.0)	64.3 (28.4)

LL, lower lumbar; *ROM*, range of motion; *UL*, upper lumbar.

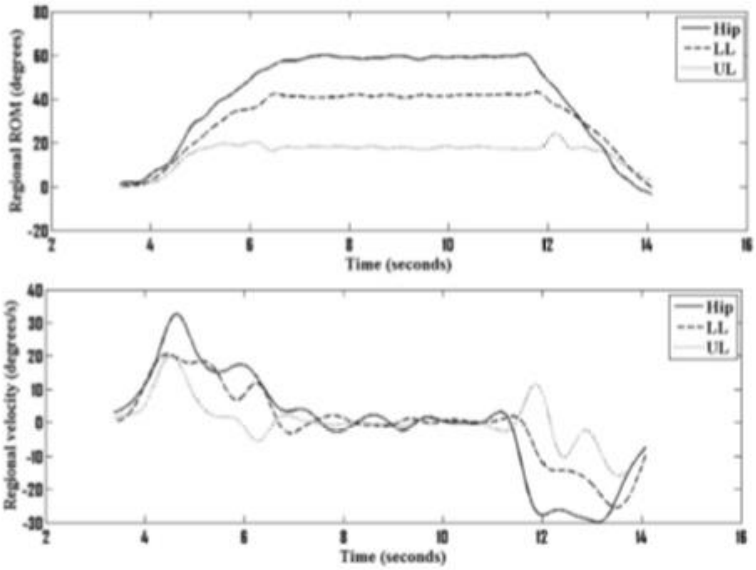

Fig 2. *ROM-time and velocity-time graphs of hip, LL, and UL during flexion task of individual participant.*

object (ie, wooden box with handles weighing 3 kg) from the floor and returning to a standing position, moving from stand-to-sit on a stool and then returning to standing.

Data Analysis

Data were captured at 30 Hz. Upper lumbar spine kinematics were derived from the relative sagittal angle between the T12 and L3 sensors and LL spine from the relative angle between the L3 and S1 sensors. Hip kinematics were derived from the relative angle between the S1 and thigh sensors. Positive and negative velocity of the upper spine, lower spine, and hip were obtained for all tasks by differentiating the ROM data. All data were normally distributed. Correlations between tasks were explored comparing ROM and velocity profiles using Pearson correlation coefficient calculated in matrix laboratory software (Matlab R2013a, MathWorks, Inc, Natick, MA). One-way analysis of variance was performed using the Statistical Package for the Social Sciences software (Statistics 20, IBM, Armonk, NY) to determine if significant differences are evident between the task kinematics. Post hoc analysis was carried out using the Tukey procedure to determine the location of any differences. Statistical significance was accepted at the 5% level for all tests.

Results

Mean (SD) ROM across all the tasks for each anatomical region is displayed in Table 1, and a single participant's ROM-time and velocity-time graphs are presented in Figure 2 for the movement of flexion. The ROM used during flexion was significantly different from those for stand-to-sit and sit-to-stand for all anatomical regions. Differences in ROM between flexion and lifting were observed for the hip only (Table 2).

Moderate to good correlations were observed between flexion and lifting for all anatomical regions investigated (r = 0.57-0.83). Moderate to good correlations were also evident between flexion and stand-to-sit for all anatomical regions (r = 0.52-0.70) as well as for flexion and sit-to-stand (r = 0.55-0.73) (Table 3).

Mean (SD) velocity across all tasks for each anatomical region is displayed in Table 1, and the differences between flexion and lifting velocity (positive and negative) were

Table 2. *Significant Differences (P value) for ROM and Velocity for Each Anatomical Region*

		Significant Difference		
		ROM (degrees)	Positive Velocity (degrees/s)	Negative Velocity (degrees/s)
UL flexion	Lifting	.206	.129	.421
	Stand-to-sit	<.001	<.001	<.001
	Sit-to-stand	<.001	<.001	.007
LL flexion	Lifting	.545	.084	.017
	Stand-to-sit	<.001	.063	<.001
	Sit-to-stand	<.001	<.001	.552
Hip flexion	Lifting	<.001	<.001	<.001
	Stand-to-sit	<.001	<.001	.990
	Sit-to-stand	<.001	.039	<.001

LL, lower lumbar; *ROM*, range of motion; *UL*, upper lumbar.

Table 3. *Correlation (r) for ROM and Velocity for Each Anatomical Region*

	Correlation		
	ROM (degrees)	Positive Velocity (degrees/s)	Negative Velocity (degrees/s)
UL flexion vs lifting	0.57	0.25	0.39
UL flexion vs stand-sit	0.52	0.16	0.06
UL flexion vs sit-stand	0.55	0.19	0.03
LL flexion vs lifting	0.83	0.29	0.53
LL flexion vs stand-sit	0.70	0.19	0.29
LL flexion vs sit-stand	0.73	0.28	0.55
Hip flexion vs lifting	0.58	0.47	0.55
Hip flexion vs stand-sit	0.67	0.24	0.31
Hip flexion vs sit-stand	0.66	0.09	0.51

LL, lower lumbar; *ROM*, range of motion; *UL*, upper lumbar.

evident for the hip and LL spine but not for the UL spine. Differences between flexion and stand-to-sit were observed for positive and negative velocity in the UL spine, as well as differences in negative velocity in the LL spine and positive velocity for the hip (Table 2). Flexion velocity was significantly different from sit-to-stand velocity at the UL spine (positive and negative) as well as for the LL spine (positive velocity) and hip (negative velocity).

Poor to moderate correlations were evident between flexion velocity and lifting velocity for all anatomical regions (r = 0.25-0.55), suggesting a limited relationship between the 2 movements. Poor to moderate correlations were also observed between flexion velocity and velocity during stand-to-sit and sit-to-stand (r = 0.03-0.55), further suggesting a limited relationship between flexion velocity and velocity used during the other functional tasks.

Discussion

This study explored the relationships between the different sagittal tasks commonly assessed within the clinical environment to determine if the resultant kinematics represent distinctly different movements. This was achieved using a novel sensor string enabling multiple anatomical regions to be studied.

The results show that, on the whole, sagittal kinematics of the hip and lumbar spine during forward flexion tasks are different from those observed during other functional tasks. This finding suggests that the movement of flexion is distinctly unique to the other movements investigated.

It is commonplace for clinicians to assess flexion in a routine clinical examination of the spine; however, these findings suggest that it may be necessary to assess other functional tasks as kinematic inferences about other movements are unlikely to be accurately drawn from assessment of flexion alone.

The results of the study show that there are similarities between flexion and lifting. At both lumbar regions, there was no difference in the ROM; the magnitude of difference was less than 2°. This suggests that participants used as much spinal flexion during lifting as they did during forward bending. Individuals seemed not to routinely alter their lumbar curvature during low load lifting, a finding observed previously within the literature.[8,16,17] Range of motion was different at the hip for lifting, where a greater range of hip flexion was used to achieve the lift. This shift in hip contribution did not seem to affect the lumbar spine, suggesting that individuals who use more hip flexion during lifting do not necessarily decrease their lumbar flexion ROM. Velocity demonstrated some distinct differences between the 2 movements for the LL and hip regions. Therefore, despite the ROM being similar, suggesting similar kinematic profiles, it is the higher order kinematics (velocity) where differences exist, demonstrating that lifting resulted in greater velocity at the LL spine and hip. Although this finding has been reported previously, it suggests that providing an individual with a target or focus to the motion seems to result in greater velocity.[8] This is a factor warranting further exploration but may have implications when interpreting the effects of lifting within the clinical setting.

Correlation, as opposed to testing for difference, explores the relationship between the ROM across the tasks (rather than the difference in ROM for each task), and the results suggest only a moderate relationship in the ROM used. A strong correlation between flexion and lifting was noted for the LL spine, suggesting a good relationship between the magnitudes of motion demonstrated between these 2 motions. This provides further evidence for the similarity in behavior between these motions for the LL region. It is not known whether an alteration in 1 of these movement profiles will directly affect the other and is something for further investigation. Only moderate correlations were noted for the UL and hip regions, providing evidence of a weaker relationship and illustrating a lack of similarity between these tasks for these regions. Therefore, caution is advised if extrapolating flexion kinematics to those of lifting for the UL and hip region.

Stand-to-sit and sit-to-stand appear to use different kinematic profiles for all anatomical regions. Compared with flexion, less spinal ROM is evident with a greater contribution provided by the hips. These findings are supported by previous studies on both lumbar flexion and sit-to-stand and stand-to-sit. [11] Furthermore, this study found a greater contribution from the LL spine during both flexion and sit-to-stand and stand-to-sit. Previous studies have explored the relative motion between the lumbar regions during sit-to-stand only,[16,18] and therefore, this study expanded the analysis to other functional tasks on lumbar regions and hip.

The inclusion of these functional tasks during clinical assessment will explore the different relationships between the lumbar spine and hip and is likely to provide different information about overall movement behavior of the lumbar-hip region than flexion alone. Self-selected velocity for flexion compared with sit-to-stand and stand-to-sit provides further evidence of the uniqueness of these tasks. Flexion was consistently completed using greater velocity for the spinal regions, compared to sit-to-stand and stand-to-sit, with the opposite being true for the hip. Velocity during flexion seems to poorly correlate with velocity used during other functional tasks, suggesting that each task has distinct properties relating to dynamic movement behavior. The correlations between velocity of different tasks for the lumbar spine and hip have not been previously explored in the literature; therefore, this novel finding provides new insights into the relationship between flexion and other tasks. Velocity has been shown to be a key determinant of movement smoothness and therefore provides important information regarding kinematics.[8] Therefore, clinically, the interrelationship between hip movement velocity and lumbar velocity cannot be fully explored using flexion alone.

This study suggests that the motion of flexion is unique in its kinematic profile. This suggests that clinicians should not be overreliant on the interpretation of flexion ROM within the clinic to determine the degree of impairment. The results suggest that other sagittal tasks are unique in how they challenge the lumbar spine and hip, and therefore, clinicians should be cautious about inferences made from assessing flexion alone. The failure to assess other movements functionally relevant to the patient is likely to result in an incomplete understanding of the movement profile. An assessment incorporating other functional tasks, even if they are in the same movement plane, may be necessary to better understand the movement behavior of these regions.

Limitations

This study was limited to a young male population. It is not known whether these results would be replicated with other ages or female participants. The population was healthy and therefore serves as a reference for an asymptomatic population[19]; however, the extrapolation of the results to those with pathology or pain may not be possible. This study focused on sagittal movements, as these are common in daily living; however, the relationship between other cardinal plane spinal motions and their functional counterparts is not known.

Further research could extend the analysis to females or differing age groups. It may be possible that, due to age-related changes in the spine, the relationship between cardinal movements and functional movements are altered. Furthermore, a similar method could be used to explore whether treatment-induced gains in ROM (eg, flexion) have any automatic effect on other more functional sagittal tasks.

Conclusion

This study suggests that sagittal tasks use different lumbar-hip kinematics and place different demands on the spine and hip. Strong correlations were only evident for the LL spine ROM between lifting and flexion; all other tasks afforded moderate or weak correlations. Significant differences were evident in the ROM and velocity comparing flexion to other sagittal tasks. These findings suggest that clinicians should not extrapolate findings from clinical testing of flexion to other functional tasks, as they demonstrate functionally unique kinematics.

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No funding sources or conflicts of interest were reported for this study.

Contributorship Information

Concept development (provided idea for the research): J.M.W., R.S.A.

Design (planned the methods to generate the results): R.S.A., M.D.J., P.S.T., J.M.W.

Supervision (provided oversight, responsible for organization and implementation, writing of the manuscript): M.D.J., J.M.W., P.S.T.

Data collection/processing (responsible for experiments, patient management, organization, or reporting data): R.S.A., J.M.W.

Analysis/interpretation (responsible for statistical analysis, evaluation, and presentation of the results): R.S.A., M.D.J., J.M.W.

Literature search (performed the literature search): R.S.A., J.M.W.

Writing (responsible for writing a substantive part of the manuscript): R.S.A.

Critical review (revised manuscript for intellectual content, this does not relate to spelling and grammar checking): J.M.W., M.D.J., P.S.T.

Practical Applications

- The study findings demonstrate that trunk flexion results in a greater or similar ROM to the other tasks investigated for both spinal regions but less ROM for the hip.
- Strong correlations for ROM are reported between flexion and lifting for the LL spine ($R = 0.83$) and all regions during stand-to-sit and sit-to-stand ($R = 0.70$-0.73).
- No tasks were strongly correlated for velocity ($R = 0.03$-0.55).
- Thus, the results show that sagittal kinematics of the hip and lumbar spine during trunk flexion are different from those observed during other functional tasks.
- This finding suggests that the movement of flexion is distinctly unique to the other movements.

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Paper-three

Investigating the contribution of the upper and lower lumbar spine, relative to hip motion, in everyday tasks

Raee S. Alqhtani [a], Michael D. Jones [a, *], Peter S. Theobald [a], Jonathan M. Williams [b]

[a] Bioengineering Research Group, School of Engineering, Cardiff University, CF24 3AA, UK
[b] Faculty of Health and Social Sciences, Bournemouth University, UK

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ABSTRACT

Background: It is commonplace for clinicians to measure range of motion (ROM) in the assessment of the lumbar spine. Traditional single 'joint' models afford measuring only a limited number of regions along the spine and may, therefore, over-simplify the description of movement. It remains to be determined if additional, useful information can be gleaned by considering the traditional 'lumbar region' as two regions.
Objective: The aim of this study was to determine whether modelling the lumbar spine as two separate regions (i.e. upper and lower), yields a different understanding of spinal movement relative to hip motion, than a traditional single-joint model. This study is unique in adopting this approach to evaluate a range of everyday tasks.
Method: Lumbar spine motion was measured both by being considered as a whole region (S1 to T12), and where the lumbar spine was modelled as two regions (the upper (L3-T12) and lower (S1-L3)).
Results: A significant difference was evident between the relative contribution from the lower and upper spine across all movements, with the lower lumbar spine consistently contributing on average 63% of the total ROM. A significant difference was also evident between the whole lumbar spine-hip ratio, and the lower lumbar spine-hip ratio, for the movement of lifting only. The lower lumbar spine achieved greater velocity for all tasks, when compared to the upper lumbar spine.
Conclusion: This study has consistently demonstrated differences in the contribution of the upper and lower spinal regions across a range of everyday tasks; hence, it would appear that greater focus should be given to performing more detailed assessments to fully appreciate spinal movement.

1. Introduction

Measuring lumbar range of motion (ROM) is typically performed using 2 sensors or markers, one at each end of the lumbar spine. This includes technologies relying on electromagnetics (Shum et al., 2005, 2007), inertial sensors (Ha et al., 2013; Williams et al., 2013) and fibre-optics (Williams et al., 2010). Calculating the resultant angle between these 2 sensors provides an estimate of lumbar range of motion, with the lumbar spine modelled as a single 'joint'. The lumbar spine, however, consists of many segments or 'joints' (L1-S1) and thus this single joint model may result in lost information about more regional lumbar spine movement behaviour. Whilst previous authors have suggested that the upper and lower lumbar spines display differences in their kinematic behaviour (Williams et al., 2012; Parkinson et al., 2013; Williams et al., 2013), traditional single 'joint' models would fail to identify such subtleties and may, therefore, over simplify the description of movement. Significant scope exists to better understand and appreciate the relationship between lumbar spine and hip kinematics, given how it both underpins rehabilitation programmes (Lee and Wong, 2002) and is associated with various forms of functional disabilities, which may have a serious impact on an individual's quality of life (Cox et al., 2000).

The dominant functional tasks such as flexion, extension, lifting and transiting from stand-to-sit or sit-to-stand have long been associated with spinal disorders and spinal pain (McGill, 1997; Dempsey, 1998). Spine and hip kinematics are closely coordinated when performing many daily tasks (Mayer et al., 1984; Pearcy et al., 1985; Strand and Wie, 1999), suggesting that lumbar spine-hip

* Corresponding author.
E-mail addresses: AlqhtaniRS@cardiff.ac.uk (R.S. Alqhtani), JonesMD1@Cardiff.ac.uk (M.D. Jones).

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disorders may affect functional tasks as well as the cardinal movements often employed in the clinic. Indeed, sit-to-stand and stand-to-sit activities are very regular daily tasks (Lomaglio and Eng, 2005), performed 60 times per day on average by working people (Dall and Kerr, 2010). The most important task that influences lumbar and hip kinematics is lifting objects from the floor, which is a common daily activity particularly amongst those working in jobs involving physical labour (Shum et al., 2005).

A series of studies have previously focused on quantifying the relationship between the lumbar spine relative to hip motion, during everyday tasks (Paquet et al., 1994; Lee and Wong, 2002; Wong and Lee, 2004; Shum et al., 2005; Shum et al., 2007); however, in all cases the lumbar spine was only considered as a single region. More recently, authors have adopted multi-regional lumbar spine models across clinical populations (Williams et al., 2012, 2013) and healthy subjects (Leardini et al., 2011; Parkinson et al., 2013), identifying differences in regional contribution. No study has yet, however, considered a multi-regional lumbar spine model versus hip motion, across a series of everyday tasks. Such data would significantly assist in achieving a better understanding of lumbar spine kinematics, especially when supplemented by multi-regional velocities (Shum et al., 2010), as the relative movement behaviour of the hip and its interaction with the lumbar spine has been suggested as being important (Lee and Wong, 2002; Sahrmann, 2002; O'Sullivan, 2005). Clinical studies have previously confirmed differences in this ratio between those with and without back pain (Shum et al., 2005, 2007), whilst alterations in this ratio affect the bending and compressive stresses on the lumbar spine (Dolan and Adams, 1993; Tafazzol et al., 2014).

Subsequently, this study investigated how the upper and lower lumbar regions contributed to spinal movement – relative to hip motion, when performing a range of everyday tasks. Comparison was drawn both to a traditional 'single-joint' measuring method, and to previous studies evaluating a single, everyday tasks (i.e. sit-to-stand).

2. Methods

2.1. Participants

Fifty-three male participants were recruited from Cardiff University (age = 29.4 ± 6.5 years; mass = 75.3 ± 16.4 kg; height = 1.69 ± 0.15 m). No participants had a history of lower extremity problems or spinal pain, surgery, rheumatological or neurological disorders. All participants provided written informed consent prior to data collection. The study was approved by the Cardiff School of Engineering Ethics Committee.

2.2. Instrumentation

Data describing lumbar spine and hip kinematics were collected using four tri-axial accelerometers (THETAMetrix, Waterlooville, UK), each with a 24 mm^2 footprint. Each sensor was then placed, using double-sided tape, over the spinous processes of S1, L3, T12 and the lateral aspect of the right thigh, mid-way between the lateral epicondyle and greater trochanter on the iliotibial band (ITB) (Fig. 1). Each accelerometer provided axial acceleration data pertaining to absolute orientation (tilt), with respect to gravity. Sensors were wired together in a 'daisy chain' arrangement and connected to a PC, running data collection software via USB. Data were captured at 30 Hz using the supplied 3A sensor software (THETA-Metrix, Waterlooville, UK), and stored for retrospective processing. This system has been found previously to have excellent repeated-measures reliability relating to spinal movement analysis, with the intraclass correlation coefficient ranging from 0.88 to 0.99, and a standard error of measurement ranging from 0.4° to 5.2° (Alqhtani et al., 2015).

2.3. Procedure

Participants' height and weight were determined prior to sensor attachment. Participants completed a warm up exercise, which included flexion, extension and rotation of the trunk, and then a period of sensor familiarisation for the participants. Prior to starting the actual trial, participants were asked to do one trial to familiarise themselves with the experimental procedure. Each participant stood barefoot on assigned markers and focused on a wall marker, set at a height of 2 m, with arms relaxed by their side. Participants were asked to complete forward bending, backward bending, lifting an object (wooden box with handles weighing 3 kg) from the floor and returning to a standing position, moving from stand to sit on a stool and then returning to standing. No further instructions on how to move were provided.

2.4. Data analysis

Raw data were transferred to MATLAB (MathWorks Inc, Natick, MA) and filtered at 6 Hz (low-pass, Butterworth) to remove high frequency noise (Scholz et al., 2001). Sagittal plane absolute angles for each sensor were determined, with respect to gravity and regional ROM was defined as the relative motion between adjacent distal and proximal sensors (relative angles). The whole lumbar spine was defined as the relative angle between the S1 and T12 sensors. The upper lumbar spine (ULS) was defined as the relative angle between the T12 and L3 sensors, and lower lumbar spine (Mills et al., 2007) as the relative angle between the L3 and S1 sensors. As the whole lumbar spine consists of six spinal joints and the ULS and LLS only three spinal joints, the regions were normalised per segment (i.e. the WLS kinematics divided by six and ULS and LLS kinematics divided by three). This normalisation enabled comparisons between the regions to be possible. The kinematics of ROM was determined as relative angle across time and angular velocity calculated by 5-point differentiation of the ROM-time data (Williams et al., 2013). The ratios of lumbar-to-hip motion for each region (ULS, LLS and WLS) were determined for each task. Therefore, the dependent variables for this study were ROM, peak velocity (negative and positive) and lumbar-hip ratio.

As this study aimed to evaluate the contribution of ULS and LLS relative to hip motion, an ANOVA was used to test for differences between the WLS, ULS and LLS (SPSS ver. 20). Post–hoc analysis was applied using the Tukey procedure to determine the location of any differences. Statistical significance was accepted at the 5% level for all tests.

3. Results

3.1. ROM

The mean (SD) ROM (normalised per segment) are presented in Table 1.

There was a significant difference in the ROM displayed by the ULS compared with the WLS for flexion, lifting and sit-to-stand (Table 2). Significant differences were also present between the LLS and WLS for flexion and lifting (Table 2).

A significant difference was evident between the relative contribution from the LLS and ULS across all movements (Table 2), with the lower lumbar spine consistently contributing on average 63% of the total ROM (Fig. 2).

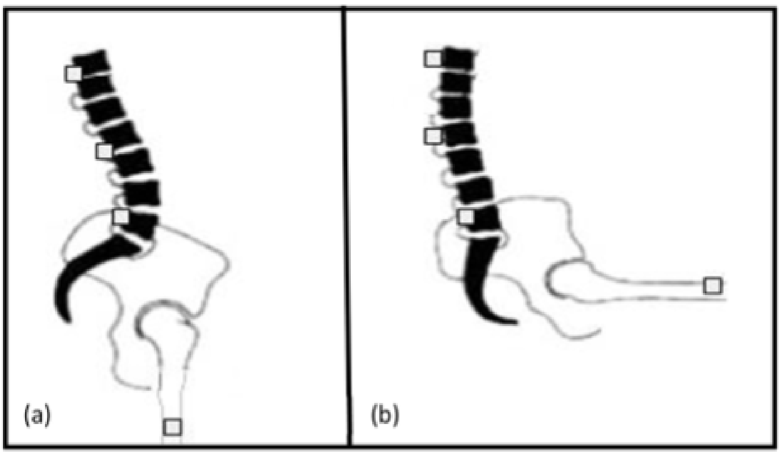

Fig. 1. Schematic representation of the three spinal sensors, over processes T12, L3 and S1. A fourth sensor was placed on the lateral aspect of the thigh, midway between the lateral epicondyle and greater trochanter on the iliotibial band (ITB). (a) Standing, (b) sitting.

Table 1
Contribution of the whole (WLS), upper (ULS) and lower (LLS) lumbar spinal regions. Range of motion data it presented as the normalised mean (degrees), with the standard deviation in parentheses.

Tasks	WLS/6	ULS/3	LLS/3
Flexion	9.8 (2.4)	7.7 (3.4)	12.0 (4.4)
Extension	−4.1 (2.6)	−2.8 (3.5)	−5.6 (4.3)
Lifting	9.3 (2.7)	7.2 (3.3)	11.8 (4.6)
Stand-to-sit	7.3 (2.8)	5.6 (3.3)	9.0 (4.9)
Sit-to-stand	7.3 (3.1)	5.4 (3.4)	8.9 (4.9)

Table 2
The p-values describing the statistical difference in range of motion data between the ULS, LLS and WLS segments (UL = 3, LL = 3 and WL = 6 segments) for each anatomical region, during a range of tasks. Statistical significance defined as $p < 0.05$, with significant data identified using an*.

Segments	Flexion	Extension	Lifting	Stand-to sit	Sit-to stand
ULS/3 vs LLS/3	<.001*	<.001*	<.001*	<.001*	<.001*
ULS/3 vs WLS/6	.006*	.191	.009*	.070	.037
LLS/3 vs WLS/6	.006*	.058	.002*	.073	.109

3.2. Ratio

The mean (SD) peak hip-lumbar ratio per segment ROM is displayed in Table 3.

A significant difference was evident between the WLS-hip ratio and the LLS-hip ratio for the movement of lifting only. No differences were noted for the WLS-hip and ULS-hip ratio. There were significant differences between the ULS-hip and LLS-hip ratio for all movements except extension (Table 4).

3.3. Velocity

Mean (SD) peak velocity for each spinal region is presented in Table 5.

A significant difference was evident between the WLS and LLS peak velocity, but only for flexion. There were significant differences between the ULS and WLS for peak velocity for stand-to-sit and lifting. No other tasks demonstrated 'per segment' peak velocity differences. Significant differences were determined between the ULS and LLS for peak velocity during all tasks, with the exception of positive velocity during extension and negative velocity during lifting (Table 6). The LLS achieved greater velocity for all tasks when compared to the ULS with the magnitude of difference ranging from 37% to 63% greater (Fig. 2).

4. Discussion

This study used a novel methodology to investigate the ratio of normalised lumbar motion, relative to hip motion. The results demonstrate few differences between each of the WLS, ULS and LLS versus hip motion, suggesting that either model may be effective in exploring lumbar spine-hip ratios. Previous studies have explored lumbar spine-hip ratios using a WLS model, with some reporting slightly higher ratios for sit-to-stand and stand-to-sit (Shum et al., 2005). Furthermore, our data indicates a proportionally greater WLS contribution to extension (than the hip) as compared to other studies (Lee and Wong, 2002; Wong and Lee, 2004), which may be due to different patient characteristics or due to a lower mean age, resulting in greater lumbar flexibility as displayed by the differences in lumbar extension ROM (Lee and Wong, 2002; Wong and Lee, 2004).

Despite the lack of difference between the WLS, and the combined ULS–LLS models, there were differences between the ULS and LLS that suggest the relationships between the hip and these specific lumbar regions are functionally different and unique. LLS-hip ratios were consistently higher than the ULS-hip ratios, due to the greater LLS ROM. This suggests that the relationship between the separate regions of the lumbar spine and hip were not equivocal and should be explored individually to appreciate the differences in kinematic behaviour.

The calculation of ratios in this manner provides insight only to the relationship of the terminal ranges, not the through range phases. Angle–angle plots can provide a description of where the ROM of each region is plotted against one another, thereby revealing further insights into kinematic behaviour. Fig. 3 illustrates the WLS plotted against the hip and the ULS-hip and LLS-hip plots for comparison (the straight-line represents a 1:1 ratio for comparison). If a WLS model was used, the behaviour would demonstrate that the hip and WLS move at a similar time and rate throughout the movement phase i.e. broadly correlating with the

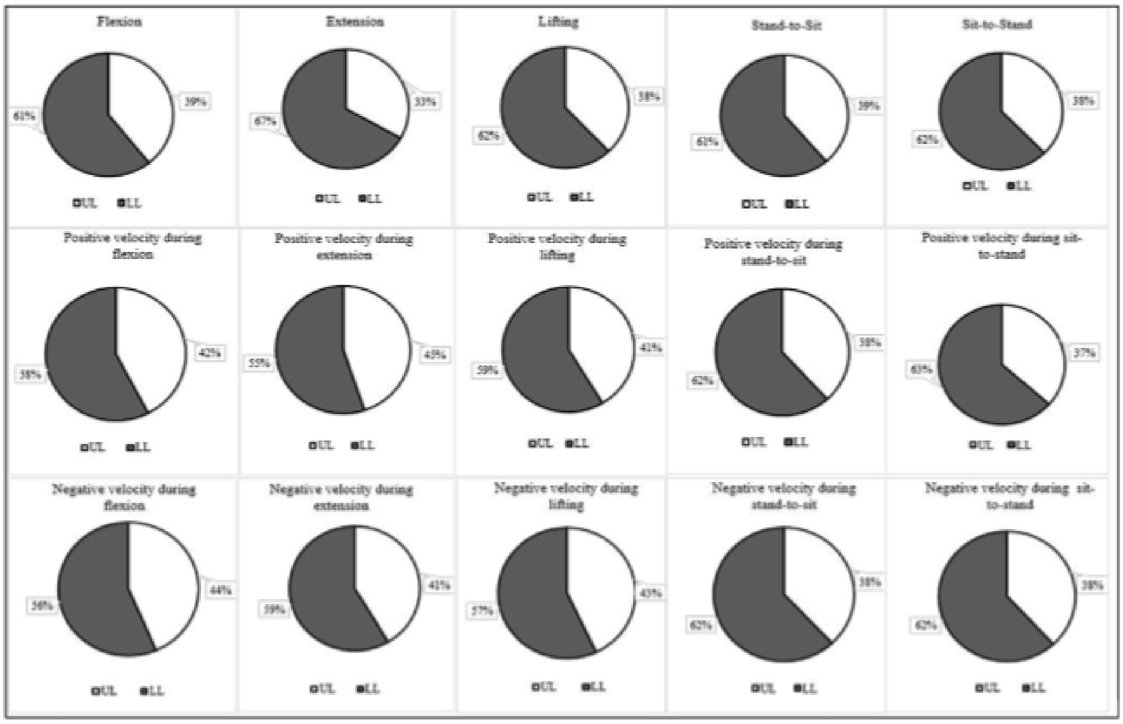

Fig. 2. Relative contributions of the upper lumbar (UL) and lower lumbar (LL) spinal regions during a series of everyday tasks.

Table 3
The ratio of normalised ROM data for ULS, LLS and WLS, versus hip ROM data (degrees). The standard deviation is presented in parentheses.

Tasks	(WLS/6)/Hip	(ULS/3)/Hip	(LLS/3)/Hip
Flexion	0.2 (0.1)	0.16 (0.1)	0.25 (0.1)
Extension	0.3 (1.5)	0.2 (1.1)	0.5 (2.3)
Lifting	0.16 (0.1)	0.1 (0.1)	0.2 (0.1)
Stand-to-sit	0.1 (0.1)	0.1 (0.1)	0.16 (0.1)
Sit-to-stand	0.1 (0.1)	0.1 (0.1)	0.16 (0.1)

Table 4
A statistical evaluation of the differences in ratio per segment for normalised ROM to hip ROM. Statistical significance defined as $p < 0.05$, with significant data identified using an*.

Segments	Flexion	Extension	Lifting	Stand-to sit	Sit-to stand
(ULS/3)/Hip vs (LLS/3)/Hip	<.001*	.556	<.001*	.004*	<.002*
(WLS/6)/Hip vs (ULS/3)/Hip	.093*	.910	.077	.234	.154
(WLS/6)/Hip vs (LLS/3)/Hip	.093*	.809	.041*	.234	.260

aforementioned straight grey line; however, the regional breakdown shows a significantly greater contribution from the hip, especially in the early phase of the motion for the LLS, where it is more even in the same phase of motion for the ULS. Such behaviour would not be visible with a WLS model.

The findings from the current study suggests that regional breakdown of the lumbar spine is also important regarding velocity. Differences between the WLS and regional spinal models were detected, as were differences between the LLS and ULS. This suggests that the ULS and LLS are functionally different for the higher order kinematics also. The velocities determined in this study were slightly greater than those reported in other studies for movements at natural speeds (Shum et al., 2005; Williams et al., 2013). These differences may be due to differences in characteristics of the sample, such as age (younger in the current study), sex (male in the

Table 5
Velocity per segment (degrees/s) of ULS, LLS and WLS segments during four tasks.

Tasks velocities		WLS/6	ULS/3	LLS/3
Flexion	+ve vel	8.6 (2.8)	7.5 (2.9)	10.5 (4.7)
	−ve vel	8.3 (3.4)	7.4 (3.3)	9.6 (4.5)
Extension	+ve vel	5.4 (3.0)	4.9 (3.0)	6.1 (4.3)
	−ve vel	4.6 (2.8)	3.9 (2.9)	5.5 (4.1)
Lifting	+ve vel	10.0 (3.4)	8.4 (3.9)	10.5 (4.7)
	−ve vel	9.3 (3.1)	7.3 (3.3)	9.6 (4.5)
Stand-to-sit	+ve vel	9.7 (3.3)	5.5 (2.5)	9.0 (4.9)
	−ve vel	5.9 (3.4)	3.3 (1.4)	5.4 (3.2)
Sit-to-stand	+ve vel	4.3 (2.2)	3.1 (1.9)	5.5 (3.5)
	−ve vel	7.5 (3.3)	5.6 (2.8)	9.2 (5.0)

Table 6
The significant difference (p-value) considering velocity between the ULS, LLS and WLS segments for each anatomical region during four tasks. Statistical significance defined as $p < 0.05$, with significant data identified using an*.

Segments		Flexion	Extension	Lifting	Stand-to sit	Sit-to stand
+ve velocity						
ULS/3	LLS/3	<.001*	.228	.021*	<.001*	<.001*
	WLS/6	.246	.771	.110	<.001*	.082
LLS/3	WLS/6	.019*	.602	.779	.600	<.060*
−ve velocity						
ULS/3	LLS/3	.011*	.039*	.091	.001*	.001*
	WLS/6	.421	.535	.029*	<.001*	.054
LLS/3	WLS/6	.218	.346	.919	<.637	<.067

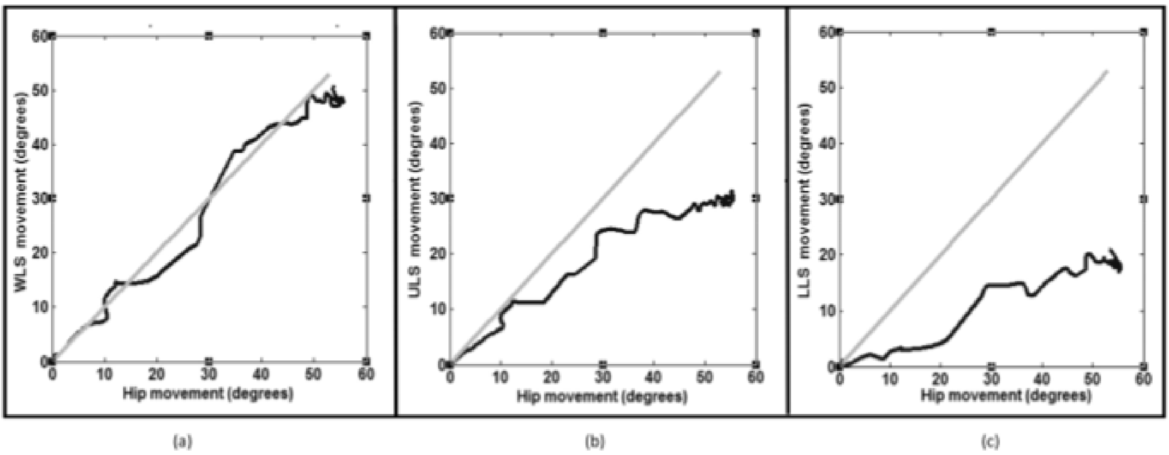

Fig. 3. The phase relationship of the lumbar spine to hip movement, with the grey line representing a sustained 1:1 relationship. These figures represent the movement phase of the WLS (a), ULS (b) and LLS (c), relative to the hip.

current study) and the presence of pain (Williams et al., 2013). The additional information gained from the regional breakdown of the lumbar spine identified that the LLS consistently moved at greater velocities. The authors are not aware of this having previously reported in the literature, though this is important since it suggests a non-even split of velocity throughout the ULS and LLS, a finding masked by a traditional single segment model.

This study also investigated the contribution of the ULS and LLS regions, relative to hip motion, when performing a range of everyday tasks. Modelling the lumbar spine as two distinct regions identified differences in the normalised (i.e. per-segment) ROM, with the LLS contribution greater than ULS by at least 2.4°, and WLS by 1.5°, over all tasks. ROM percentages per LLS segment were greater than ULS over all five tasks (Fig. 2). Hence, it was evident that modelling the WLS underestimated the LLS motion by as much as 37%, and over-estimated the ULS motion by as much as 45%. Whilst this finding is in agreement with previous studies (Williams et al., 2010; Leardini et al., 2011; Williams et al., 2012; Parkinson et al., 2013; Williams et al., 2014), the current study was the first to adopt a method of normalisation to enable a quantitative comparison. The findings are consistent with studies adopting stereoradiography (Pearcy et al., 1985) and cadaveric testing (Yamamoto et al., 1989), contributing to the increasing body of evidence that suggests a non-uniform breakdown of ROM contribution for the lumbar segments. Subsequently, this indicates that simply modelling the lumbar spine as a whole region may omit some important kinematic information, and under-estimate the LLS contribution.

The findings of the current study have important clinical ramifications. Clinicians are beginning to advocate the assessment of two separate functional regions within the lumbar spine (O'Sullivan, 2005; Dankaerts et al., 2006), with the belief that these are functionally individual. This study confirms that indeed there are functional differences in the ROM of lumbar spine models, and velocity of motion during a range of functional tasks and provides support for the use of a more detailed spinal kinematic model. Greater contributions to motion from the lower lumbar spine, as well as greater movement velocities, may help to explain increased prevalence of low back pain or pathological change in this spinal region more than the upper lumbar (Biering-Sørensen, 1983; Beattie et al., 2000). Usually, greater degeneration takes place in the lower lumbar spinal segments (Twomey and Taylor, 1987; Quack et al., 2007) and it is assumed that this is due to greater mechanical stress upon this region (Adams and Hutton, 1983). Assessment of the lumbo-pelvic rhythm has also been suggested during clinical assessment of the back (O'Sullivan, 2005), as the hip motion effects the resultant bending stresses (Dolan and Adams, 1993) and muscle activities, as well as the forces acting on the lumbar spine (McGill et al., 2000; O'Sullivan et al., 2002; Kamińska et al., 2010). Insights into lumbo-pelvic rhythm can be afforded through the determination of ratios and angle—angle plots, and this study provides novel detail regarding the regional spinal ratios.

This study provides further evidence for the separation of the whole lumbar spine into smaller regional sections, as suggested previously (Parkinson et al., 2013), to truly determine detailed kinematic information for the lumbar spine.

Limitations of the current study include a single sex population preventing the extrapolation of the findings to females. The sample was representative of a young non-impaired population and findings relating to more elderly, or those in pain or impaired, may differ from the current findings. Analysis was limited to the sagittal plane and more detailed 3-dimensional kinematics would provide detail regarding out of plane motions.

5. Conclusion

The findings of the current study suggest modelling the lumbar spine as two distinct regions demonstrates normalised kinematic differences compared to treating the lumbar spine as a whole. It is evident that modelling the lumbar spine as a whole entity underestimates the contribution from the LLS and over-estimates the contribution from the ULS. This suggests that to model the lumbar spine as a whole may omit some important kinematic information. Clinicians should be aware of the differences between the regions to better inform their clinical assessment of the lumbar spine.

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Conference Abstracts

Conference abstract 1

Alqhtani R, Jones M, Theobald P and Williams J. 2014. A novel method to evaluate the viability of 3A sensor measurements of primary motions for six cephalo-caudal regions and demonstrate range of motion for each particular region in 3D. *In: International Conference on Spinal Manipulation* 25-27 October 2013 Phoenix, USA.

A novel method to evaluate the viability of 3A sensor measurements of primary motions for six cephalo-caudal regions and demonstrate range of motion for each particular region in 3D

Background:

Practitioners must justify their choice of treatment modality based on an objective assessment of range of motion (ROM). The plan and decisions regarding intervention and treatment of the spine are often partially dependent on joint ROM. Physiotherapists need to test ROM before and after the session to demonstrate the effectiveness of treatment. Therefore, the practitioners need a valid, reliable and portable device to measure inter-segmental spine ROM rather than measuring only the three main regions (i.e. cervical, thoracic and lumbar). Researchers and practitioners have used various means to measure head-cervical and spinal ROM with either invasive or non-invasive measurement techniques. Many factors such as setting, cost or radiation hazard have limited using a range of these pieces of equipment both clinically and in research. A non-invasive, portable method (3A sensors) were obtained to examine its feasibility on specific spinal kinematics as well as to demonstrate ROM of inter-segmental areas (six cephalo-caudal regions) in 3D. These regions were the head-cervical (HC), upper thoracic (UThx), middle thoracic (MThx), lower thoracic (LThx), upper lumbar (ULx) and lower lumbar (LLx).

Objectives:

The purpose of this study was to investigate the reliability of a novel motion analysis device for measuring the regional breakdown of spinal motion and describing the relative motion of different segments of the thoracolumbar (TL).

Methods:

Two procedures were used in this study.

Procedure one: One sensor was placed on the subject's forehead and another on T1. In a seated position, the participant moved their neck in flexion, extension and right and left lateral bending. In a prone position, the participant rotated his neck right and to left.

Procedure two: Six sensors were placed on the spinous processes of T1, T4, T8, T12, L3 and S1 and the participant was instructed to move his trunk in flexion, extension, right and left lateral bending from a standing position and then rotate their trunk to the right and left when lying down on their side.

Results:

The inter-class correlation coefficients (ICC) of repeated measurements of all six regions were found to be high, ranging from 0.881 to 0.994 for all three planes. The error values ranged from 0.68º to 5.2° for all regions in all directions. Flexion of HC, UThx, MThx, LThx, ULx and LLx were 66°, 3°, 3°, 36°, 19° and 15° respectively. Extension of HC, UThx, MThx, LThx, ULx and LLx were 61°, 7°, 11º, 7°, 5º and 21° respectively. Right lateral bending of HC, UThx, MThx, LThx, ULx and LLx were 41°, 6°, 7º, 12°, 12º and 12° respectively. Left lateral bending of HC, UThx, MThx, LThx, ULx and LLx were 42°, 5°, 7º, 12°, 11º and 11° respectively. Right rotation of HC, UThx, MThx, LThx, ULx and LLx were 74°, 9°, 34º, 21°, 6º and -14° respectively. Left rotation of HC, UThx, MThx, LThx, ULx and LLx were 80°, -11°, 26º, 22°, 5º and 8° respectively.

Conclusion:

It was concluded that the 3A sensors represent a viable and accurate device to assess spinal ROM. In addition, the findings of this study demonstrate that the ROM of each region of four vertebrae, between T1 and S1, were comparable with previous studies. This method and findings might open a window for researchers and practitioners to focus on these regions rather than measuring only the three main areas of spine (cervical, thoracic and lumbar).

Conference abstract 2

Alqhtani R, Jones M, Theobald P and Williams J. 2014. The reliability of novel multiregional spinal motion measurement device. International Journal of Therapy And Rehabilitation, 21, S6-S6.

The reliability of novel multiregional spinal motion measurement device

RS Alqhtani
MD Jones
PS Theobald
JM Williams
Institute of Medical Engineering and Medical Physics, Cardiff School of Engineering, Cardiff University

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ABSTRACT | PDF | PDF PLUS

Background:
Current spinal range of motion (ROM) measurement methods have limitations ranging from the amount of detail obtained to environmental costs and complexity. In particular, limited regional spinal motion is obtained using the current methods. However, a new portable 'string' of accelerometers is proposed to overcome these limitations.

Objectives:
This study seeks to determine the reliability of this sensor string in measuring three-dimensional spinal ROM and to investigate the relative motions across six different regions.

Methods:
Two procedures were undertaken on 18 healthy participants. Protocol one: two sensors were placed on the forehead and T1 to measure cervical ROM; and protocol two: six sensors were placed on the spinous processes of T1, T4, T8, T12, L3 and S1 to measure thoraco-lumbar regional ROM.

Results:
The ICC values for all regions were found to be high, ranging from ICC=0.88–0.99 for all movements and regions of the spine, demonstrating that the proposed methods were highly reliable for repeated measures. The standard error of the means (SEMs) were small, ranging from 0.7–5.2°. The flexion/extension motion demonstrated a mean SEM of 1.9° and 1.1° for lateral bending motions. Slightly larger SEMs were observed for rotation, especially for the upper thoracic (UT) and mid thoracic (MT) region with an overall mean SEM of 3.1°. Minimum detectable change (MDC) values ranged from 1.9–14.4°. The flexion/extension motion demonstrated a mean MDC of 5.2° with 3.1° for lateral bending motions. Slightly larger MDCs were observed for rotation (mean MDC=8.4°), especially for the UT and MT region.

Implications:
This method was able to quantify the relative contribution of differing regions to the overall motion. The method described represents a reliable method of assessing spinal ROM across multiple spinal regions.

Conference abstract 3

Alqhtani R, Jones M, Theobald P and Williams J. 2014. Hip and lumbar motion: Is there a correlation between flexion and functional tasks? International Journal of Therapy And Rehabilitation, 21, S7-S7.

Hip and lumbar motion: Is there a correlation between flexion and functional tasks?

RS Alqhtani
JM Williams
MD Jones
PS Theobald
School of Health and Social Care, Bournemouth University

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ABSTRACT | PDF | PDF PLUS

Background:
It is common for patients with spinal pain to report evoked pain, associated with a variety of everyday tasks. Despite this, the assessment of an individual with spinal pain usually involves the completion of movements in the cardinal planes, for example flexion and extension. The relationship between these cardinal motions and more functional tasks is yet to be established. The aim of this study was to explore the relationship between flexion and more functional everyday sagittal tasks.

Methods:
Fifty three participants were recruited for this study (age 29.4 years [SD=6.5], weight 75.3 kg [SD=16.4] and height 1.7 m [SD=15.4]). Four daisy-chained accelerometers were attached to the skin over the S1, L3, T12 and lateral thigh. Each sensor provided absolute orientation with respect to gravity, which was used to determine tilt in the sagittal plane. Relative angles between adjacent sensors were used to quantify the motion for the hip, lower lumbar and upper lumbar spine. Ratios between peak values were calculated by dividing one regional peak ROM by another, quantifying regional contribution to motion. Pearson correlation coefficients were used to explore the relationship between the movements of flexion, lifting, stand-to-sit and sit-to-stand.

Results:
Strong correlations for range of motion are reported between flexion and lifting for the lower-lumbar spine (R^2=0.83) and all regions during stand-to-sit and sit-to-stand (R^2=0.84–0.92). Strong correlation is also noted between flexion and lifting for hip/lower-lumbar spine ratio (0.84). No tasks were highly correlated for velocity (R^2=0.01–0.72).

Conclusions:
The lower-lumbar spine appears to demonstrate a relationship between flexion and lifting. However, with the exception of stand-to-sit and sit-to-stand, no other strong correlations between regional range-of-motion, velocity and ratio for a series of sagittal tasks were observed.

Implications:
These findings suggest that each functional task is distinct using a differing degree of range of motion, velocity and relative motion at the lumbar spine and hip. It is recommended that clinicians explore more than just the cardinal planes of motion in the examination of the spine, as these do not demonstrate the same kinematics as more functional tasks.

Prizes:

Prize for the Best Poster awarded at 33rd Phsiotherapy Research Society Spring 2014

Meeting on The 14th May 2014 at University of East Anglia

Physiotherapy Research Society

Prize for the Best Poster

awarded at the 33rd Physiotherapy Research Society
Spring 2014 Meeting on the
14th May 2014 at the University of East Anglia

To

RAEE ALQHTANI

....

Congratulations from the Physiotherapy Research Society

50UEA
Fifty Years of the University of East Anglia

Appendix B-Information and Consent sheets

Subject Information Form

Name of Researcher: Raee Al Qhtani, a PhD student at Cardiff School of Engineering
Supervisors: Dr Mike Jones and Dr Peter Theobald
Cardiff University Engineering Department,
The Parade,
Cardiff University
Title of study: Developing a Methodology to Perform Measurements of the Multi-spinal regions and Lumbar-Hip Complex Kinematics during Dominant Daily Tasks
Equipment:
3A Pearl Sensors (tri-axial accelerometer) connected to a laptop by a mini USB cable, double-sided hypoallergenic tape and a chair.
Invitation Paragraph
You have been invited to participate in a research study carried out by the Institute of Medical Engineering and Medical Physics (IMEMP), Cardiff University. Before you decide whether you would like to participate, please take a few minutes to read this information sheet so that you gain a better understanding of what the research involves. If you have any questions please do not hesitate to contact the research team or discuss the information on this sheet with others. Please take your time deciding whether you would like to participate.
Before you participate, the researchers will give you an opportunity to ask any questions you might have.
Thank you for taking the time to read this.
What is the purpose of the study?
The purpose of this study is to investigate into the movement behaviour of the spine in the sagittal plane using the 3A Pearl Sensors (tri-axial accelerometer).
Do I have to take part?

You have no obligation to take part and participation is entirely voluntary. If you do decide to participate you will be required to sign a Participant Consent Form. You are free to withdraw from the study at any time with giving a reason.

What will happen to me if I take part?

After reading the information sheet, the researcher will give you the opportunity to ask any questions regarding this study and will ask you to sign a consent form. You will need to take off your shirt and the door of the study room will be closed to maintain privacy.

Procedure: Six sensors will be attached to specific areas on your forehead, back and thigh using double-sided hypoallergenic tape. From a standing position, you will be asked to move your torso forwards, backwards, lifting object, standing to sitting and sitting to standing.

What if something goes wrong?

If you are harmed in taking part in the research there are NO specific compensation arrangements. If you are harmed due to someone's negligence your legal rights are unaffected.

What are the possible risks?

There is no anticipated risk from participating in this study. However, if an unforeseen situation does occur there is a qualified first aider available in same building.

Who will know I am taking part in the study?

The researcher and you only know about your participation in this study.

Will my information be kept confidential?

Yes, all information is kept strictly confidential. Any information obtained and used in papers/presentations etc will only refer to the subjects as numbers/letters and not by name or any other reference.

What will happen to the results of the study?

The results of the study will be used as a reliability study which will be a part of a PhD thesis, conference/journal papers, and conference presentations and posters. As stated above, any personal information will remain confidential and subjects will be referred to by number/letter only.

Who is organising the trials?

The trials have been organised by Raee Alqhtani, a PhD researcher. The supervisors of the trial are Dr Mike Jones and Dr Peter Theobald. Contact details can be found at the bottom of this information sheet.

What if I wish to lodge a complaint?

If have a question about any aspect of this study, you should speak to Raee Alqhtani from Cardiff University who is organising this study. If you remain unhappy and wish to complain formally, you can do this by contacting Dr Mike Jones (jonesmd1@cf.ac.uk) or Dr Theobald theobaldps@cf.ac.uk who are in charge of this study.

Who has reviewed this study?

This study has been reviewed and approved by Cardiff University Ethics committee.

What do I do now?

Thank you for considering taking part in this trial. If you are still happy to take part in the trial please sign the consent form attached.

Further information
If you have any other questions please do not hesitate to contact Raee Alqhtani for more information.
Contact Details:
Mr. Raee Alqhtani
Cardiff School of Engineering
Cardiff University Queen's Building
The Parade
Cardiff
CF24 3AA
Email: alqhtani@cf.ac.uk

Consent Form

Name of Researcher*:* Raee Al Qhtani, a PhD student at Cardiff School of Engineering
Supervisors: Dr Mike Jones and Dr Peter Theobald.
Cardiff University Engineering Department,
The Parade,
Cardiff University
Title of study: ***Developing a Methodology to Perform Measurements of the Multi-spinal regions and Lumbar-Hip Complex Kinematics during Dominant Daily Tasks***

Please Initial box
I confirm I have read and understood the information sheet, dated …/…. /........ ☐
I have had the opportunity to consider the information and ask any questions. ☐
Any questions I have asked have been answered to my satisfaction. ☐
I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason. ☐
I give permission for the researcher to photograph the procedure for data analysis purposes. ☐
I understand that all information about me will be kept in confidential and destroyed after the stipulated time period. ☐ ☐
I agree to take part in this study. ☐

Name of particepant……………………….. …………………………………..
Signature ……………………………………………….………... Date …………..
Name of Witness (Researcher) ………………………………………………
Signature ……………………………………………...….. Date

Appendix C- MatLab codes of spine-hip kinematics

Matlab codes for flexion tasks

```
%%%% for Raee %%%%
%%%% first plot one sensor graph to determine values for inputs
%%%%
%%%% flexion is positive and extension is negative
%%%% INPUTS REQUIRED ln. 23
%%%% INPUTS REQUIRES ln. 78 + 79
%%%% INPUTS REQUIRED ln. 49 + 50
%%%% INPUTS REQUIRED ln. 62 + 63
%%%% INPUTS REQUIRED ln. 85 %%%% INPUTS REQUIRED ln. 547 + 548
rad2deg = 180/pi;
%%% need to trim off the zeros at the beginning. Identify them from a plot %%% of one
sensor. Then input into here. aa =350;
%%%% define variables or arguments in %%%%
absolutepitchFemur = Sensor_6_Angle_X(aa:end)*rad2deg;
 absolutepitchS1 = Sensor_5_Angle_X(aa:end)*rad2deg;
 absolutepitchL3 = Sensor_4_Angle_X(aa:end)*rad2deg;
 absolutepitchT12 = Sensor_3_Angle_X(aa:end)*rad2deg;
 absolutepitchT1 = Sensor_2_Angle_X(aa:end)*rad2deg;
absolutepitch4head = Sensor_1_Angle_X(aa:end)*rad2deg;
 if 1
%%%%%%% deal with the pitch flip
%%%%%%% check which sensor is flipping
%%%%%%% locate point of inflection (first flip) and the end of inflection
%%%%%%% from graph
bb1 = 144; %%%taken from graph bb2 = 421; %%% taken from graph bb3 =
absolutepitchT1(bb1:bb2); bb4 = (90-bb3); bb5 = 90+bb4;
absolutepitchT1(bb1:bb2) = bb5; end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%% second flip
if 0
%%%%%%% deal with the pitch flip
%%%%%%% first flip
bb6 = 605;
%%%taken from graph
bb7 = 670;
%%% taken from graph
bb8 = absolutepitchT1(bb6:bb7);
```

```
bb9 = (90-bb8);
bb10 = 90+bb9;
if 1
ddd1= 1055; %%%taken from graph ddd2 = 1136; %%% taken from graph ddd3 =
absolutepitchT12(ddd1:ddd2); ddd4 = (90-ddd3); ddd5 = 90+ddd4;
absolutepitchT12(ddd1:ddd2) = ddd5;
end
%%%%%% check which sensor is flipping
%%%%%% locate point of inflection (first flip) and the end of inflection
%%%%%% from graph

%%%%%%filter%%%%%%%
if1
rom(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end rom(maxix) = 0; rom(maxix-1) = 0; rom = rom/0.033; %sample duration.
absolutepitchFemurfilt =rom ;
clear rom;
%%%%%%%%%filter
 [b11,a11] = butter(4,6/30,'low');
x = filtfilt(b11,a11,absolutepitchS1);
maxix =length(x);
rom(1)=0;
rom(2)=0;
for i
x = (3:maxix-2)
rom(ix) = (x(ix-2)-*x(ix8-ix+1) 1)+8*x(-x(ix+2))/12;
end
rom(maxix) = 0;
rom(maxix-1)= 0;
rom = rom/0.033;
%sample duration.
absolutepitchS1filt =rom ;
clear
rom;
%%%%%%%%%filter
b11,a11] = butter(4,6/30,['low';)
x = filtfilt(b11,a11,absolutepitchL3);
maxix =length(x);
rom(1)=0;
rom(2)=0;
for
ix = (3:maxix-2)rom(ix) = (x(ix-2)-8*x(ix-ix+1)1)+8*x(-x(ix+2))/12;
end
rom(maxix) = 0;
```

```
rom(maxix-= 0;1)rom = rom/0.033;
%sample duration.
absolutepitchL3filt =rom ;
clear
rom;
%%%%%%%%filter[b11,a11] = butter(4,6/30,'low');
x = filtfilt(b11,a11,absolutepitchT12);
maxix =length(x);
rom(1)=0;
rom(2)=0;
for
ix = (3:maxix-2) rom(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
rom(maxix) = 0;
rom(maxix-1)= 0;rom = rom/0.033;
%sample duration.
absolutepitchT12filt =rom ;
clear
rom;
%%%%%%%%filter
[b11,a11] = butter(4,6/30,'low');
x = filtfilt(b11,a11,absolutepitchT1);maxix=length(x);
rom(1)=0;  rom(2)=0;  for  ix  =  (3:maxix-2)  rom(ix)  =  (x(ix-2)-8*x(ix-1)+8*x(ix+1)-
x(ix+2))/12; end
rom(maxix) = 0;
rom(maxix-1)= 0;rom = rom/0.033;
%sample duration.
absolutepitchT1filt =rom ;
clear
rom;
%%%%%%%%filter
b11,a11] = butte[r(4,6/30,'low');
x = filtfilt(b11,a11,absolutepitch4head);
maxix =length(x);
rom(1)=0;
rom(2)=0;
for
ix = (3:maxix-2)rom(ix) = (x(ix-2)-*x(ix8-+1)1)+8*x(ix-x(ix+2))/12;
end
rom(maxix) = 0;
rom(maxix-1)= 0;
rom =
rom/0.033;
%sample duration.
```

```
absolutepitch4headfilt =rom ;
clear
rom;
%%%%%%%%%%%%
absolutepitchFemurromfilt = filtfilt(b11,a11,absolutepitchFemur);
absolutepitchS1romfilt = filtfilt(b11,a11,absolutepitchS1);
absolutepitchL3romfilt = filtfilt(b11,a11,abs
olutepitchL3);
absolutepitchT12romfilt = filtfilt(b11,a11,absolutepitchT12);
absolutepitchT1romfilt = filtfilt(b11,a11,absolutepitchT1);
absolutepitch4headromfilt = filtfilt(b11,a11,absolutepitch4head);
%%%%% plot all raw sensor data for axis of interest %
%%%
figure(1)
plot(absolutepitchFemurromfilt,'b';)
hold
on
plot(absolutepitchS1romfilt,'g';)
plot(absolutepitchL3romfilt,'r');
plot(absolutepitchT12romfilt,'c';)
hold off
title('absolute angles all sensors');
legend('AbsAngFemur','AbsAngS1','AbsAngL3','AbsAngT12','AbsAngT1','AbsAng4head'
);
%%%%%% determine motion onset %%%%%%%%%%%%%%%%%%%
%%% calculate motion onsets %%%%%%%%
%%% need to identify the relatively flat portion of the graph
aaa = 115;
% this number needs modifying
staticsensor = absolutepitc
hFemur(1:aaa);
meanstaticsensor = mean(staticsensor);
sdstaticsensor = std(staticsensor);
counter=0;
cutoffsensor = abs(3* sdstaticsensor);
%request absolute value (not neg or pos)
for n=1:length(absolutepitchFemur)
if abs(absolutepitchFemur(n)-meanstaticsensor)>= cutoffsensor counter=counter+1;
else counter=0;
break end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%
taticsensor = absolutepitchS1(1:aaa);
meanstaticsensor = mean(staticsensor);
```

```
sdstaticsensor = std(staticsensor);
counter=0;
cutoffsensor = abs(3* sdstaticsensor);
%request absolute value (not neg or poss)
for
n=1:length(absolutepitchS1)
if
abs(absolutepitchS1(n)-meanstaticsensor)>= cutoffsensorcounter=counter+1;
else
counter=0;
end
if
counter ==30
onset = n-29;
break
end
S1onset = onset
end
if
counter ==30
onset = n-29;
end
Femuronset = onset
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%
staticsensor = absolutepitchT12(1:aaa);
%%%%%%
staticsensor = absolutepitchL3(1:aaa);
meanstaticsensor = mean(staticsensor);
sdstaticsensor = std(staticsensor);
counter=0;
cutoffsensor = abs(3* sdstaticsensor);
%request absolute value (not neg or poss)
for
n=1:length(absolutepitchL3)
if
abs(absolutepitchL3(n)-meanstaticsensor)>= cutoffsensorcounter=counter+1;
else
counter=0;
end
if
counter ==30
onset = n-29;
break
```

```
end
L3onset = onset
meanstaticsensor = mean(staticsensor); sdstaticsensor = std(staticsensor); counter=0;
cutoffsensor = abs(3* sdstaticsensor); %request absolute value (not neg or poss)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%
staticsensor = absolutepitch4head(1:aaa); meanstaticsensor = mean(staticsensor);
sdstaticsensor = std(staticsensor); counter=0; cutoffsensor = abs(3* sdstaticsensor);
%request absolute value (not neg or poss) for n=1:length(absolutepitch4head) if
abs(absolutepitch4head(n)-meanstaticsensor)>= cutoffsensor counter=counter+1;
for
n=1:length(absolutepitchT12)
if
abs(absolutepitchT12(n)
-
meanstaticsensor)>= cutoffsensor
counter=counter+1;
else
counter=0;
end
if
counter ==30
onset = n-29;
break
end
end
T12onset = onset
%%%%%%%%%%%%
staticsensor = absolutepitchT1(1:aaa);
meanstaticsensor = mean(staticsensor);
sdstaticsensor = std(staticsensor);
counter=0;
cutoffsensor = abs(3* sdstaticsensor);
%request absolute value (not neg or poss)
for
n=1:length(absolu
tepitchT1)
if
abs(absolutepitchT1(n)-meanstaticsensor)>= cutoffsensorcounter=counter+1;
else
counter=0;
end
if
counter ==30
```

```
onset = n-29;
break
end
end
T1onset = onset
else counter=0; end if counter ==30; onset = n-29; break
end
end
Foreheadonset = onset
%%%%%%%%%%%%%%
zeroedabsolutepitchFemur = absolutepitchFemurromfilt - (mean(absolutepitchFemur(1:aaa))); zeroedabsolutepitchS1 = absolutepitchS1romfilt - (mean(absolutepitchS1(1:aaa))); zeroedabsolutepitchL3 = absolutepitchL3romfilt - (mean(absolutepitchL3(1:aaa))); zeroedabsolutepitchT12 = absolutepitchT12romfilt - (mean(absolutepitchT12(1:aaa))); zeroedabsolutepitchT1 = absolutepitchT1romfilt-(mean(absolutepitchT1(1:aaa))); zeroedabsolutepitch4head = absolutepitch4headromfilt - (mean(absolutepitch4head(1:aaa))); figure(2) plot(zeroedabsolutepitchFemur,'b'); hold on plot(zeroedabsolutepitchS1,'g'); plot(zeroedabsolutepitchL3,'r'); plot(zeroedabsolutepitchT12,'c'); plot(zeroedabsolutepitchT1,'m'); plot(zeroedabsolutepitch4head,'y');
hold off
title('zeroed absolute angles all sensors'); legend('AbsAngFemur','AbsAngS1','AbsAngL3','AbsAngT12','AbsAngT1','AbsAng4head');
%%%% calculate relative angle for each REGION of interest %%%% relativepitchHip = zeroedabsolutepitchS1 - zeroedabsolutepitchFemur;
 relativepitchLowerLx = zeroedabsolutepitchL3 - zeroedabsolutepitchS1; relativepitchUpperLx = zeroedabsolutepitchT12 - zeroedabsolutepitchL3;
relativepitchLx = zeroedabsolutepitchT12 - zeroedabsolutepitchS1;
relativepitchTx = zeroedabsolutepitchT1 - zeroedabsolutepitchT12;
 relativepitchCx = zeroedabsolutepitch4head - zeroedabsolutepitchT1;
 figure(3)
plot(relativepitchHip,'b'); hold on
plot(relativepitchLowerLx,'g'); plot(relativepitchUpperLx,'r'); plot(relativepitchTx,'c'); plot(relativepitchCx,'m'); hold off
title('zeroed relative angles all regions');
legend('RelAngHip','RelAngLLx','RelAngULx','RelAngTx','RelAngCx')
%%%%%% all data trains must be equal length %%%%%%
if 1
%%%%% run to complete cross correlation to calcluate lags (time delay) %%% absabsolutepitchFemur = abs(absolutepitchFemur);
 absabsolutepitchS1 = abs(absolutepitchS1);
 absabsolutepitchL3 = abs(absolutepitchL3);
 absabsolutepitchT12 = abs(absolutepitchT12);
```

```
absabsolutepitchT1 = abs(absolutepitchT1);
 absabsolutepitch4head = abs(absolutepitch4head);
[cHip, lags] = xcorr(absabsolutepitchFemur, absabsolutepitchS1,'coeff');
 [cHip, i] = max(cHip); samplelagcHip = lags(i);
[cLLx, lags] = xcorr(absabsolutepitchS1, absabsolutepitchL3,'coeff');
samplelagsall = [samplelagcHip samplelagcLLx samplelagcULx samplelagcTx samplelagcCx];
figure(4)
cLLx, i] = max(cLLx);
[samplelagcLLx = lags(i);[cULx, lags] = xcorr(absabsolutepitchL3, absabsolutepitchT12,'coeff');
[cULx, i] = max(cULx);
samplelagcULx = lags(i);
[cTx, lags] = xcorr(absabsolutepitchT12, absabsolutepitchT1,'coeff');
cTx, i] = max(cTx);
[samplelagcTx = lags(i);[cCx, lags] = xcorr(absabsolutepitchT1, absabsolutepitch4head,'coeff');
[cCx, i] = max(cCx);
samplelagcCx = lags(i);

%%%% to plot lags %%%%
Vel(maxix-1) = 0;
Vel = Vel/0.033; %sample duration.
absolutepitchS1Vel = Vel;
clear vel;
x = filtfilt(b1,a1,absolutepitchL3); maxix =length(x); Vel(1)=0;
barh(samplelagsall);
title('bar chartfor lags between sensor pairs (in samples)');
legend('lagcHip lagcLLx lagcULx lagcTx lagcCx')
%%%% calc velocity for each sensor %%%%%
%%% filter displacement data %%%%
b1,a1] = butter(4,6/30,['low');
[b2,a2] = butter(4,2/30,'low');
x = filtfilt(b1,a1,
absolutepitchFemur);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-1)= 0;
Vel = Vel/0.033;
```

```
%sample duration.
absolutepitchFemurVel = Vel;
clear
vel;
%%%%
%%%%%%
x = filtfilt(b1,a1,absolutepitchS1);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
243
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-= 0;
1)
absolutepitchFemurVelfilt            =            filtfilt(b2,a2,absolutepitchFemurVel);
absolutepitchS1Velfilt = filtfilt(b2,a2,absolutepitchS1Vel);
absolutepitchL3Velfilt = filtfilt(b2,a2,absolutepitchL3Vel);
 absolutepitchT12Velfilt = filtfilt(b2,a2,absolutepitchT12Vel);
 absolutepitchT1Velfilt = filtfilt(b2,a2,absolutepitchT1Vel);
absolutepitch4headVelfilt = filtfilt(b2,a2,absolutepitch4headVel);
 figure(5) plot(absolutepitchFemurVelfilt,'b');
hold on
plot(absolutepitchS1Velfilt,'g';)
plot(absolutepitchL3Velfilt,'r';)
plot(absolutepitchT12Velfilt,'c';)
plot(absolutepitchT1Velfilt,'m';)
plot(absolutepitch4headVelfilt,'y';)
hold off
title('absolute anglular Velocity all sensors filter';)
%%%%% velocity of each region %%%%%
x = filtfilt(b2,a2,relativepitchHip);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-*x(ix8-+1)ix1)+8*x(-x(ix+2))/12;
```

```
end
Vel(maxix) = 0;
Vel(maxix-1)= 0;
Vel = Vel/0.033;
%sample duration.
relativepitchHipVel = Vel;
clear
vel;
%%%%%%%%%%%%%%%%%%%%%%%%%
x = filtfilt(b2,a2,relativepitchLowerLx);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-1)= 0;
Vel = Vel/0.033;
%s
ample duration.
relativepitchLowerLxVel = Vel;
clear
vel;
%%%%%%%%%%
x = filtfilt(b2,a2,relativepitchUpperLx);
maxix =length(x);
Vel(1)=0; Vel(2)=0; for ix = (3:maxix-2)
Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-1) = 0;
Vel = Vel/0.033; %sample duration.
relativepitchUpperLxVel = Vel;
clear
vel;
%%%%%%%%%%%%%%%%%%%
x = filtfilt(b2,a2,relativepitchTx);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
```

```
Vel(maxix) = 0;
Vel(maxix-= 0;1)Vel = Vel/0.033;
 %sample duration.
relativepitchTxVel = Vel;
clear
vel;
%%%%%%%%%%%%%%%%%%%%%%%
x = filtfilt(b2,a2,relativepitchCx);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-*x(ix8-+1)ix1)+8*x(-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-1)= 0;
Vel = Vel/0.033;
%sample duration.
relativepitchCxVel = Vel;
clear
vel;
%%%%%%%%%%%%%%%%%
x = filtfilt(b2,a2,relativepitchLx);
maxix =length(x);
Vel(1)=0;
Vel(2)=0;
for
ix = (3:maxix-2)Vel(ix) = (x(ix-2)-8*x(ix-1)+8*x(ix+1)-x(ix+2))/12;
end
Vel(maxix) = 0;
Vel(maxix-1)= 0;
Vel = Vel/0.033;
%sample duration.
relativepitchLxVel = Vel;
clear
vel;
figure(6)
plot(relativepitchHipVel,'b');
hold
on
plot(relativepitchLowerLxVel,'g';)
plot(relativepitchUpperLxVel,'r');
plot(relativepitchTxVel,'c');
plot(relativepitchCxVel,'m'); hold off
```

```
title('relative anglular velocity all regions filtered');
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
figure(7)
subplot(2,1,1)
plot(relativepitchHip)
grid
on
title(HipROM')
subplot(2,1,2)
plot(relativepitchHipVel)
grid
on
title(
'HipVel')
figure(8)
subplot(2,1,1)
plot(relativepitchLowerLx)
grid
on
title('LLxROM')
subplot(2,1,2)
plot(relativepitchLowerLxVel)
grid
on
title('LLxVel')
figure(9)
subplot(2,1,1)
plot(relativepitchUpperLx)
grid
on
title('ULxROM')
subplot(2,1,2)
plot(relativepitchUpperLxVel)
grid
on
title('ULxVel')
figure(10)
subplot(2,1,1)
plot(relativepitchTx)
grid
on
title('TxROM')
subplot(2,1,2)
plot(relativepitchTxVel)
```

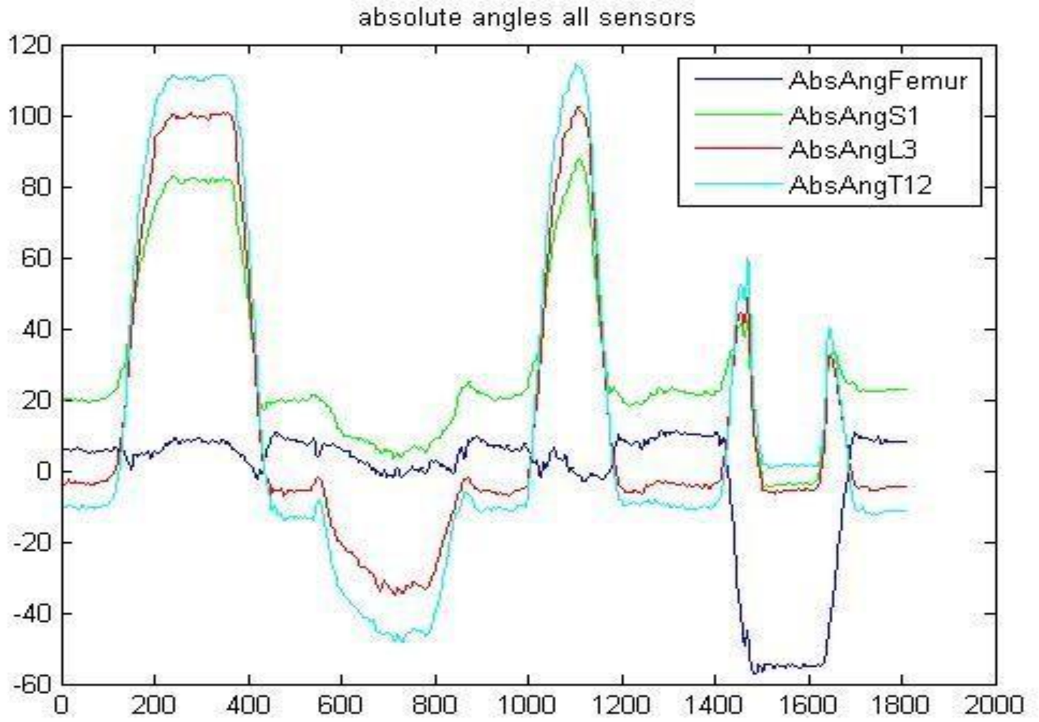
```
grid
on
title('TxVel')
figure(11)
subplot(2,1,1)
plot(relativepitchCx)
grid
on
title('CxROM')
subplot(2,1,2)
plot(relativepitchCxVel)
grid
on
title('CxVel')
%%%% To calculate absolute standing posture (i.e. from sensors) %%%%%%%
standpostabsoluteFemur = mean(absolutepitchFemur(1:(Femuronset-5)))
standpostabsoluteS1 = mean(absolutepitchS1(1:(S1onset-5))) standpostabsoluteL3 =
mean(absolutepitchL3(1:(L3onset-5))) standpostabsoluteT12 =
mean(absolutepitchT12(1:(T12onset-5))) standpostabsoluteT1 =
mean(absolutepitchT1(1:(T1onset-5))) standpostabsolute4head =
mean(absolutepitch4head(1:(Foreheadonset-5))) standspinalpostabsolute =
[standpostabsoluteS1...
standpostabsoluteL3 standpostabsoluteT12 standpostabsoluteT1]; xxxx = [1 2 3 6];
figure(12)
plot(standspinalpostabsolute,xxxx,'o') title('standing/sitting spinal posture S1-
T1:1=s1;2=L3;3=T12;4=T1'); %%%%%%% To calculate relative (regional) standing
posture standpostHip = standpostabsoluteFemur - standpostabsoluteS1 standpostLLx =
standpostabsoluteS1 - standpostabsoluteL3 standpostULx = standpostabsoluteL3 -
standpostabsoluteT12 standpostTx = standpostabsoluteT12 - standpostabsoluteT1
standpostCx = standpostabsoluteT1 - standpostabsolute4head standspinalpostrelative =
[standpostLLx standpostULx standpostTx standpostCx]; figure(13)
plot(standspinalpostrelative,xxxx,'o') title('standing/sitting spinal posture (relative):
1=LLx;2=ULx;3=Tx;4=Cx'); end
%%%%%%%%%%filtering%%%%
figure(14) subplot(2,1,1)
plot(time(104:447),relativepitchHip(104:447),time(104:447),relativepitchLowerLx(104:
447),time(
104:447),relativepitchUpperLx(104:447))
plot(time(104:447),relativepitchHipVel(104:447),time(104:447),relativepitchLowerLxV
el(104:447 ),time(104:447),relativepitchUpperLxVel(104:447))
grid
on
title('Hip ROM')
subplot(2,1,2)
```

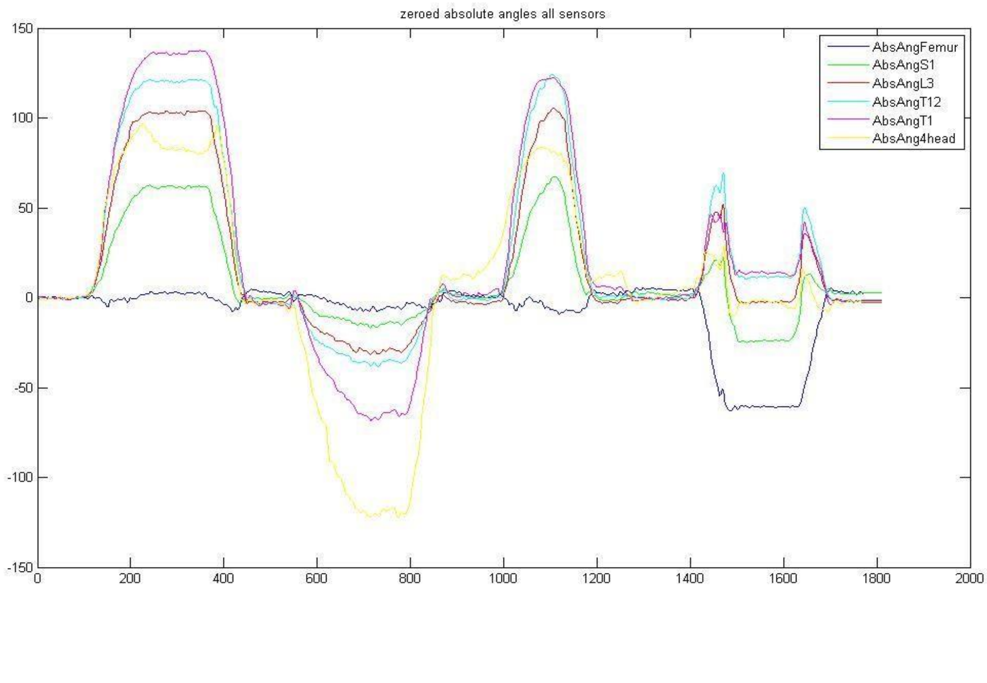
```
grid
on
title('Hip Vel')
figure(15)
subplot(2,1,1)
plot(time(104:447),relativepitchLowerLx(104:447))
grid
on
title('LL ROM')
subplot(2,1,2)
plot(time(104:447),relativepitchLowerLxVel(104:447))
grid
on
title('LL Vel')
figure(16)
subplot(2,1,1)
plot(time(104:447),relativepitchUp
perLx(104:447))
grid
on
title('UL ROM')
subplot(2,1,2)
plot(time(104:447),relativepitchUpperLxVel(104:447))
grid
on
title('UL Vel')
figure(17)
subplot(2,1,1)
plot(time(104:447),relativepitchTx(104:447))
grid
on
title('TxROM')
subplot(2,1,2)
plot(time(104:4(104:447)),relativepitchTxVel47)
grid
on
title('TxVel')
figure(18)
subplot(2,1,1)
plot(time(104:447),relativepitchCx(104:447))
grid
on
title('CxROM')
subplot(2,1,2)
```

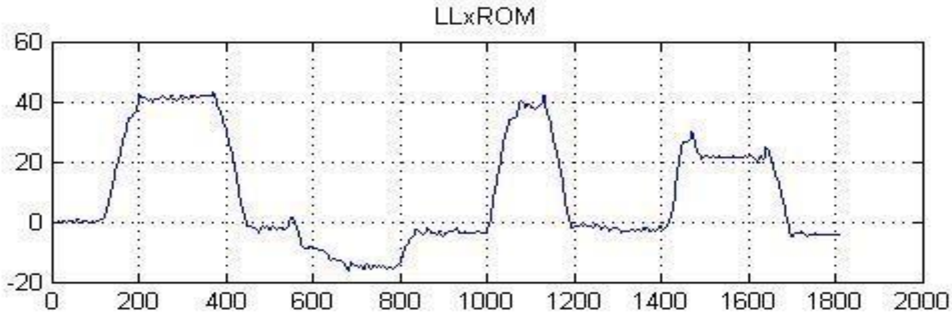
```
plot(time(104:447),relativepitchCxVel(104:447))
grid
on
title('CxVel')
%%%%%%%Phases Polts
nter
OnsetRel = 104;
PeakRel=307;
OffsetRel = 447;
figure(19)
plot(relativepitchHip(OnsetRel:PeakRel),relativepitchLowerLx(OnsetRel:PeakRel),'b')
hold
on
plot(relativepitchHip(PeakRel:OffsetRel),relativepitchLowerLx(PeakRel:OffsetRel),'r')
a = [0:1:relat
ivepitchHip(PeakRel)]
b = [0:1:relativepitchHip(PeakRel)];
plot(a,b,'g')
grid
on
hold
off
xlabel('Hip (degrees)');
ylabel('LowerLx (degrees)');
title('phase plot hip and LowerLx');
figure (20)
plot(relativepitchLowerLx(OnsetRel:PeakRel),relativepitchUppe
rLx(OnsetRel:PeakRel),'b')
hold
on
plot(relativepitchLowerLx(PeakRel:OffsetRel),relativepitchUpperLx(PeakRel:OffsetRel
),'r')
a = [0:1:relativepitchHip(PeakRel)]
b = [0:1:relativepitchHip(PeakRel)];
plot(a,b,'g')
grid
on
hold
off
xlabel('LowerLx (degrees)');
ylabel('UpperLx (degrees)');
title('phase plot LowerLx and UpperLx';)
figure (21)
plot(relativepitchUpperLx(OnsetRel:PeakRel),relativepitchTx(OnsetRel:PeakRel),'b')
```

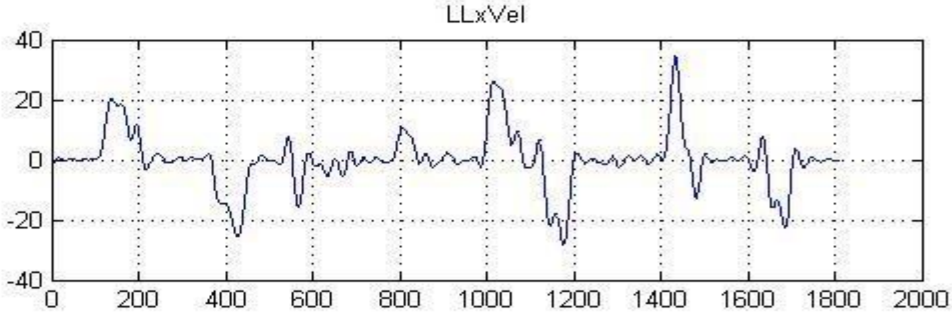
```
hold
on
plot(relativepitchUpperLx(PeakRel:OffsetRel),relativepitchTx(PeakRel:OffsetRel),'r')
a = [0:1:relativepitchHip(PeakRel)]
b = [0:1:relativepitchHip(PeakRel)];
plot(a,b,'g')
grid
on
hold
off
xlabel('UpperLx (degrees)');
ylabel('Tx (degrees)';
title('phase plot UpperLx and Tx');
figure (22)
plot(relativepitchTx(OnsetRel:PeakRel),relati
vepitchCx(OnsetRel:PeakRel),'b')
hold
on
plot(relativepitchTx(PeakRel:OffsetRel),relativepitchCx(PeakRel:OffsetRel),'r')
a = [0:1:relativepitchHip(PeakRel)]
b = [0:1:relativepitchHip(PeakRel)];
plot(a,b,'g')
grid
on
hold
off
xlabel('Tx (degrees)');
ylabel('Cx (degrees)';
title('phase plot Tx and Cx');
figure(23)
subplot(2,1,1)
plot(time(104:447),relativepitchLx(104:447))
grid
on
title('LxROM')
subplot(2,1,2)
plot(time(104:447),relativepitchLxVel(104:447))
grid
on
title('LxVel'
```

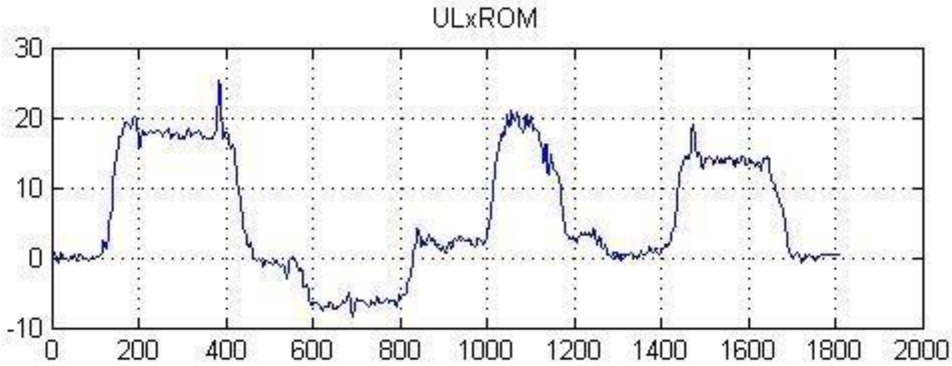
Examples of MatLab figures which have used for data analysis

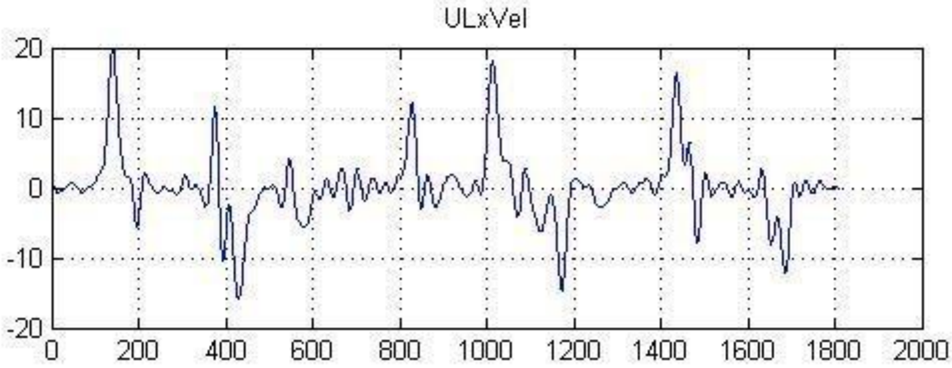


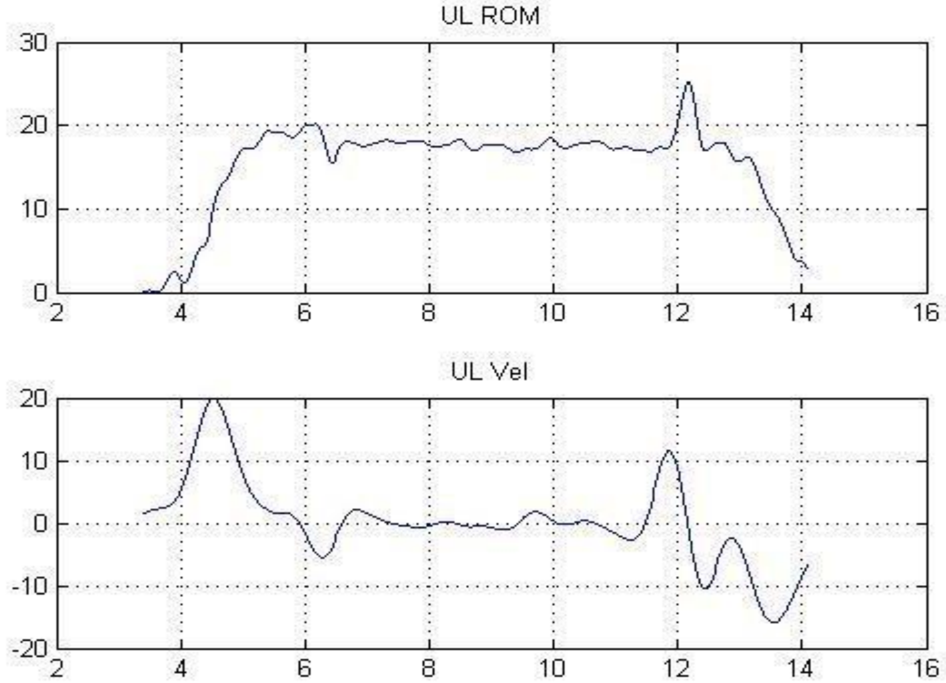


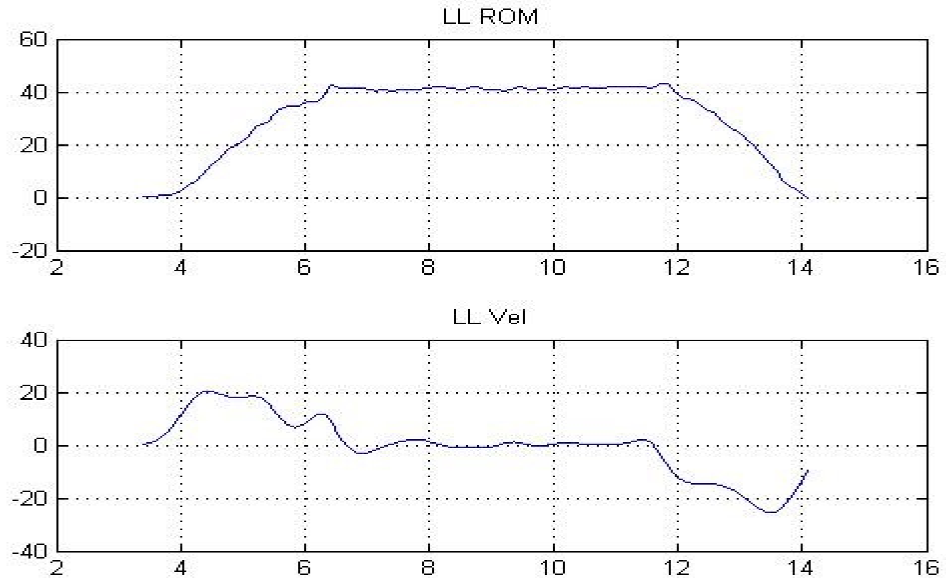

LLxROM
60
40
20
0
-20
0 200 400 600 800 1000 1200 1400 1600 1800 2000

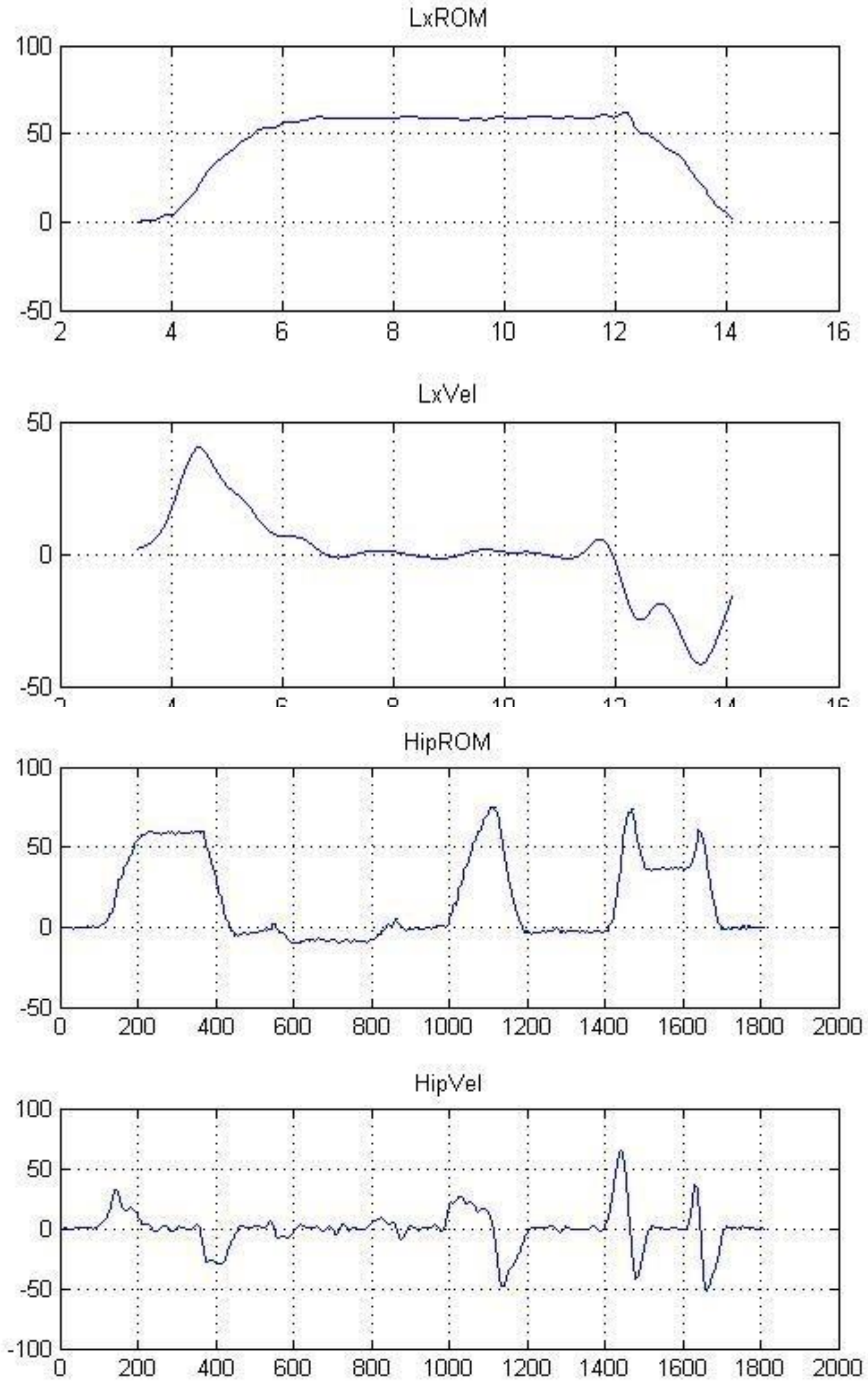

LLxVel
40
20
0
-20
-40
0 200 400 600 800 1000 1200 1400 1600 1800 2000

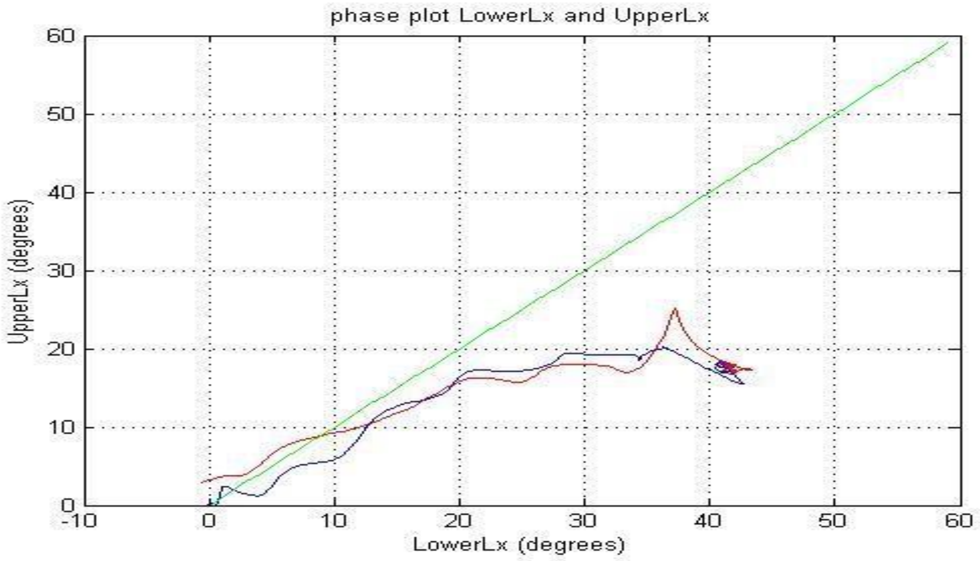

ULxROM
30
20
10
0
-10
0 200 400 600 800 1000 1200 1400 1600 1800 2000


ULxVel
20
10
0
-10
-20
0 200 400 600 800 1000 1200 1400 1600 1800 2000


UL ROM
0
10
20
30
2
4
6
8
10
12
14
16
UL Vel
-20
-10
0
10
20
2
4
6
8
10
12
14
16


LL ROM
-20
0
20
40
60
2
4
6
8
10
12
14
16
LL Vel
-40
-20
0
20
40
2
4
6
8
10
12
14
16


LxROM
100
50
0
-50
2
4
6
8
10
12
14
16
LxVel
50
0
-50
HipROM
100
50
0
-50
0
200
400
600
800
1000
1200
1400
1600
1800
2000
HipVel
100
50
0
-50
-100
0
200
400
600
800
1000
1200
1400
1600
1800
2000


phase plot LowerLx and UpperLx
UpperLx (degrees)
LowerLx (degrees)
60
50
40
30
20
10
0
-10
0
10
20
30
40
50
60


phase plot hip and LowerLx
LowerLx (degrees)
Hip (degrees)
60
50
40
30
20
10
0
-10
-10
0
10
20
30
40
50
60
70

Appendix D- Tests for Normal Distribution and homogeneity of variance

i. **Age, weight and height normal distribution**

Reliability group

Tests of Normality[b]

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	Df	Sig.
Age	.139	18	.200*	.940	18	.288
weight	.293	18	.000	.741	18	.000
height	.211	18	.033	.900	18	.056

a. Lilliefors Significance Correction

*. This is a lower bound of thne true significance.

b. group = 1.00

Case Processing Summary[a]

	Cases					
	Valid		Missing		Total	
	N	Percent	N	Percent	N	Percent
age	18	100.0%	0	.0%	18	100.0%
weight	18	100.0%	0	.0%	18	100.0%
height	18	100.0%	0	.0%	18	100.0%

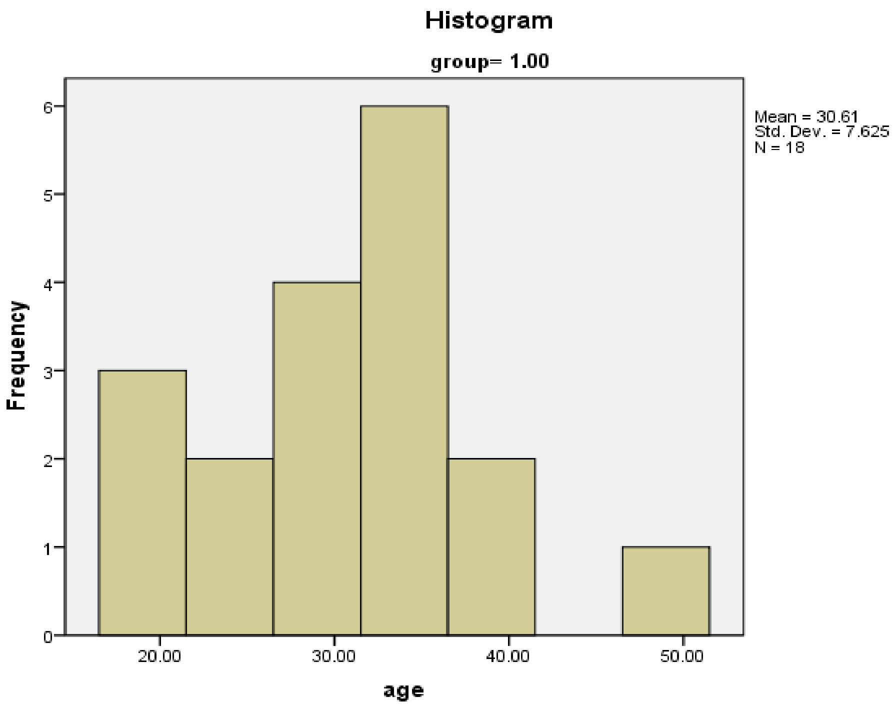
a. group = 1.00

Descriptives[a]

			Statistic	Std. Error
Age	Mean		30.6111	1.79713
	95% Confidence Interval for Mean	Lower Bound	26.8195	
		Upper Bound	34.4027	
	5% Trimmed Mean		30.3457	
	Median		32.0000	
	Variance		58.134	
	Std. Deviation		7.62456	
	Minimum		20.00	
	Maximum		43.00	
	Range		28.00	
	Interquartile Range		11.75	
	Skewness		-.004	.536
	Kurtosis		-.084	1.038
Weight	Mean		76.5556	3.41203
	95% Confidence Interval for Mean	Lower Bound	69.3568	
		Upper Bound	83.7543	
	5% Trimmed Mean		74.9506	
	Median		72.0000	
	Variance		209.556	
	Std. Deviation		14.47603	
	Minimum		65.00	
	Maximum		117.00	
	Range		52.00	
	Interquartile Range		11.00	
	Skewness		1.906	.536
	Kurtosis		3.156	1.038
Height	Mean		170.8889	1.25216
	95% Confidence Interval for Mean	Lower Bound	168.2471	
		Upper Bound	173.5307	
	5% Trimmed Mean		171.2099	
	Median		170.0000	
	Variance		28.222	
	Std. Deviation		5.31246	
	Minimum		156.00	
	Maximum		180.00	
	Range		24.00	
	Interquartile Range		3.75	
	Skewness		-.897	.536
	Kurtosis		2.968	1.038

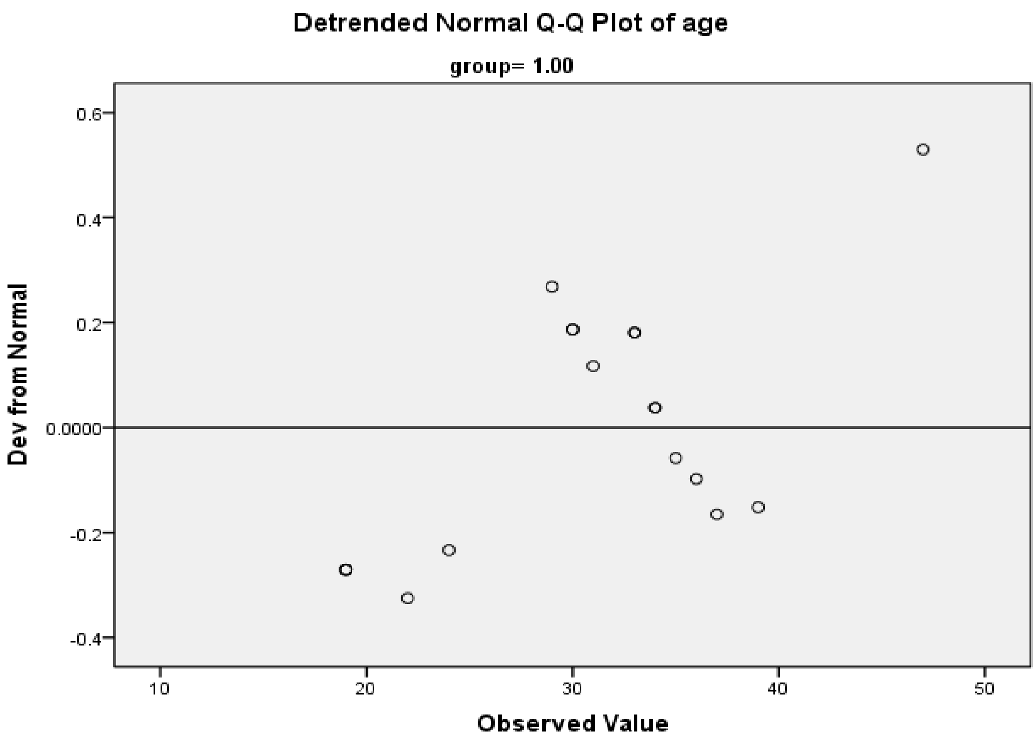
a. group = 1.00

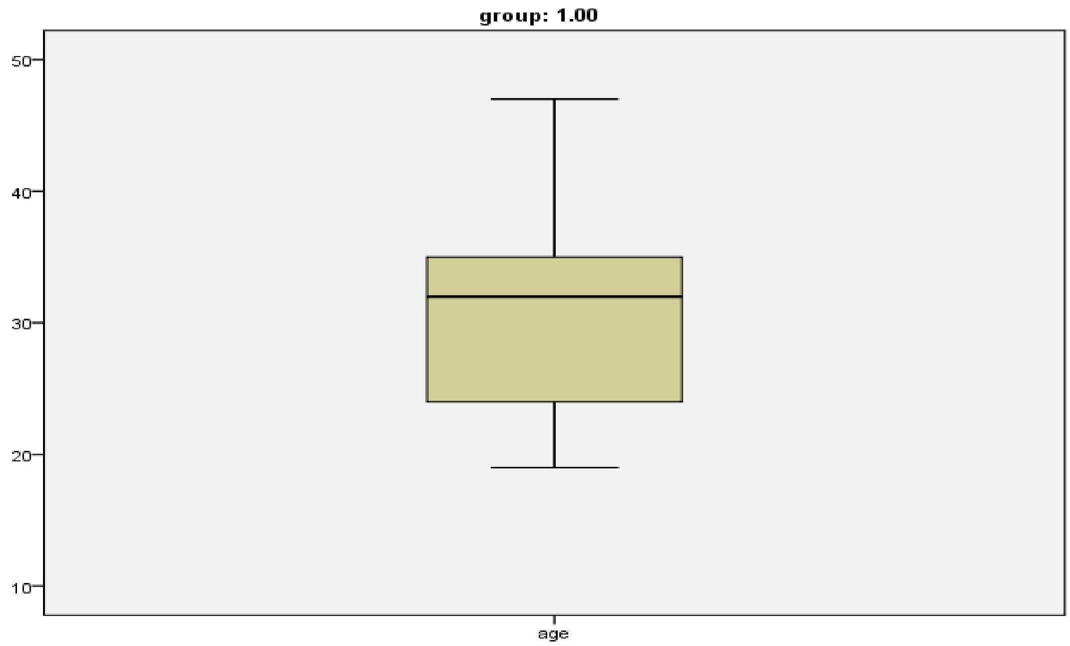
Age



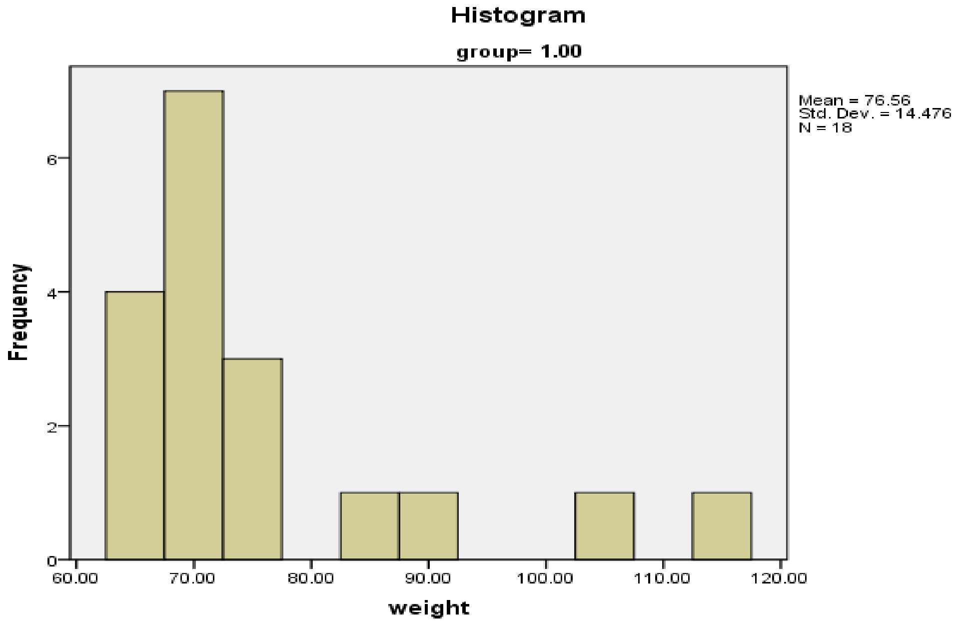
Histogram
group= 1.00
Mean = 30.61
Std. Dev. = 7.625
N = 18
Frequency
0
1
2
3
4
5
6
20.00
30.00
40.00
50.00
age

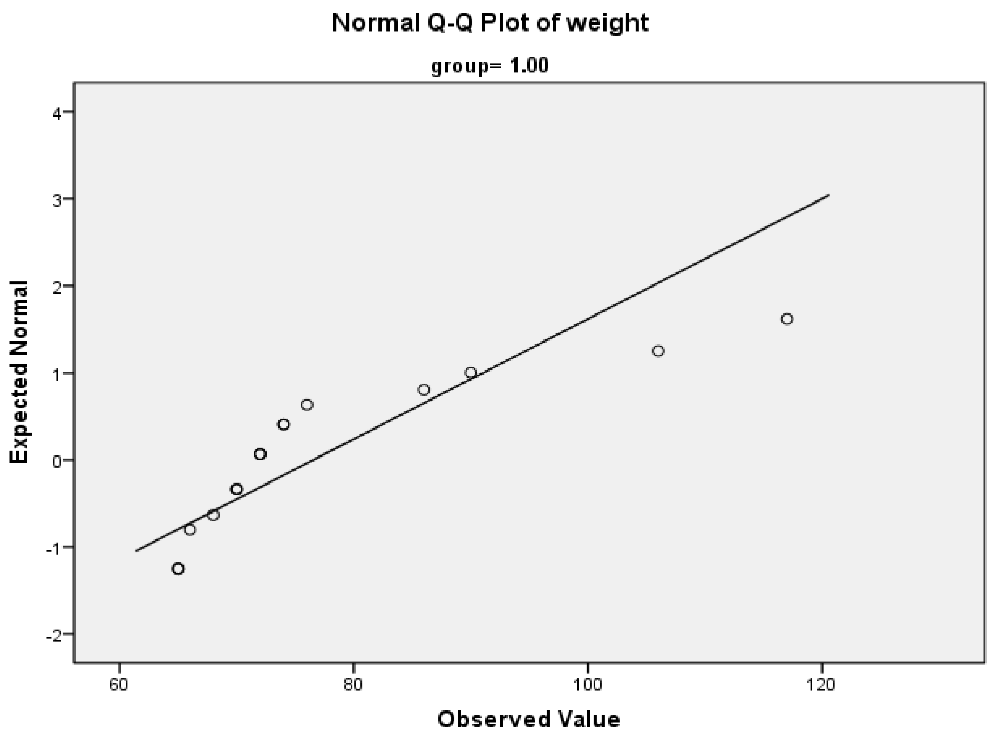

Normal Q-Q Plot of age
group= 1.00
Expected Normal
3
2
1
0
-1
-2
10
20
30
40
50
Observed Value

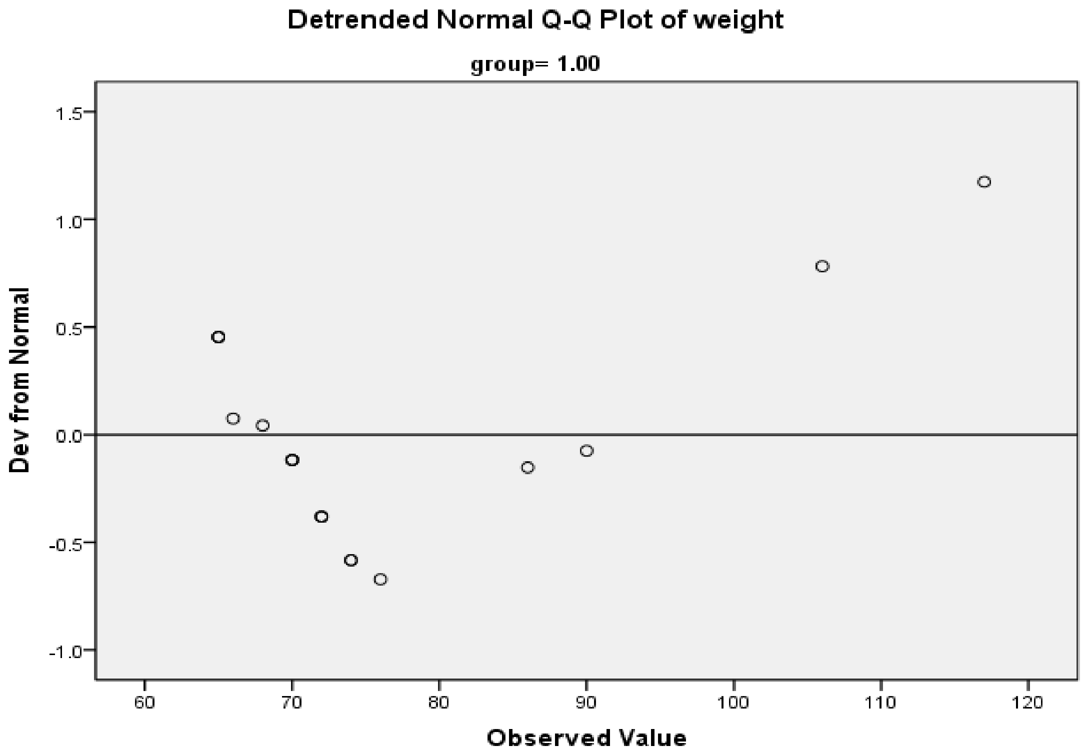

Detrended Normal Q-Q Plot of age
group= 1.00
Dev from Normal
0.6
0.4
0.2
0.0000
-0.2
-0.4
10
20
30
40
50
Observed Value

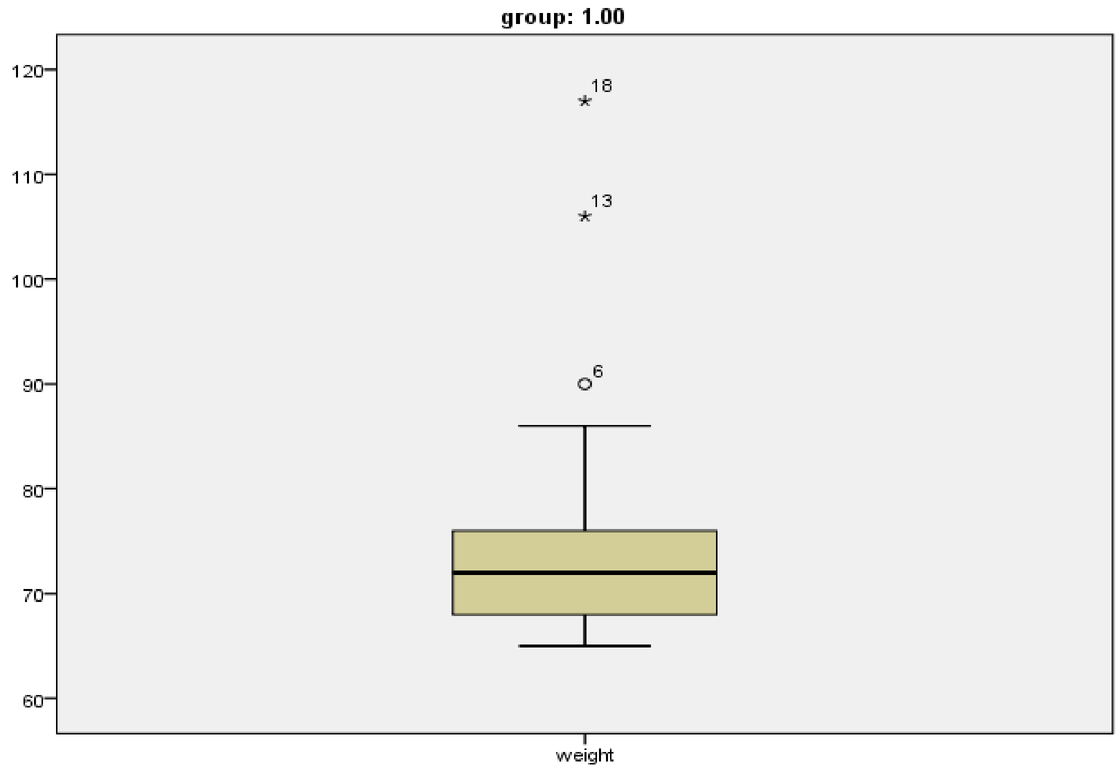

group: 1.00
50
40
30
20
10
age

Weight

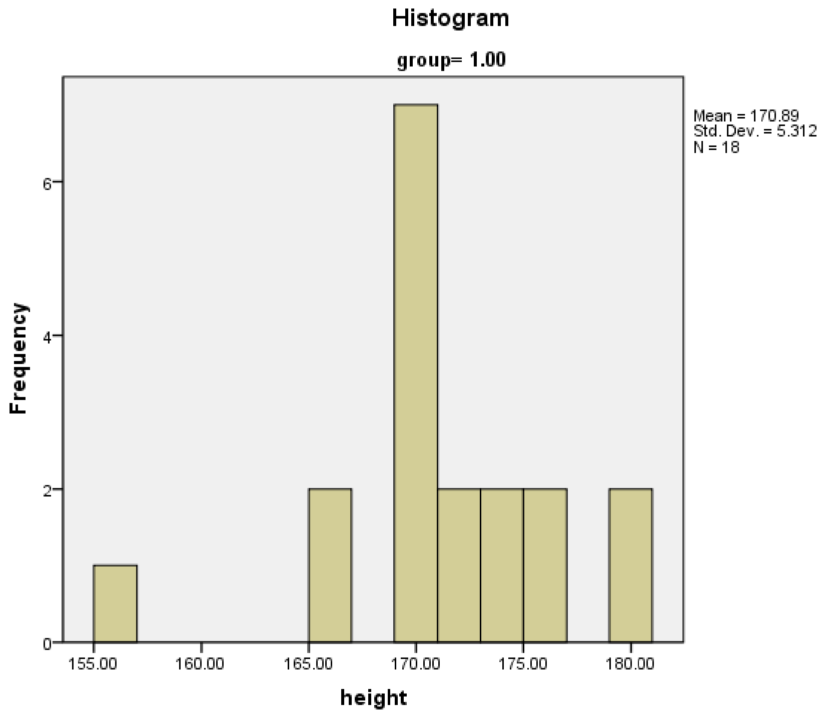

Histogram
group= 1.00
Mean = 76.56
Std. Dev. = 14.476
N = 18
Frequency
0
2
4
6
60.00
70.00
80.00
90.00
100.00
110.00
120.00
weight

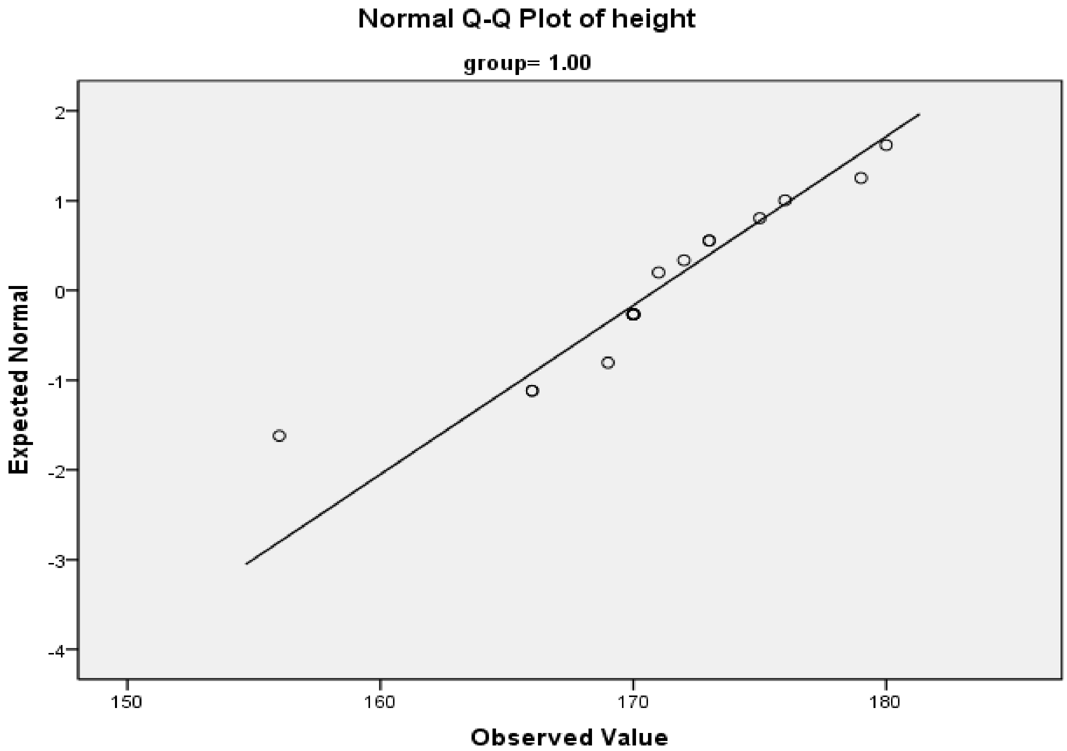

Normal Q-Q Plot of weight
group= 1.00
Expected Normal
Observed Value
4
3
2
1
0
-1
-2
60
80
100
120

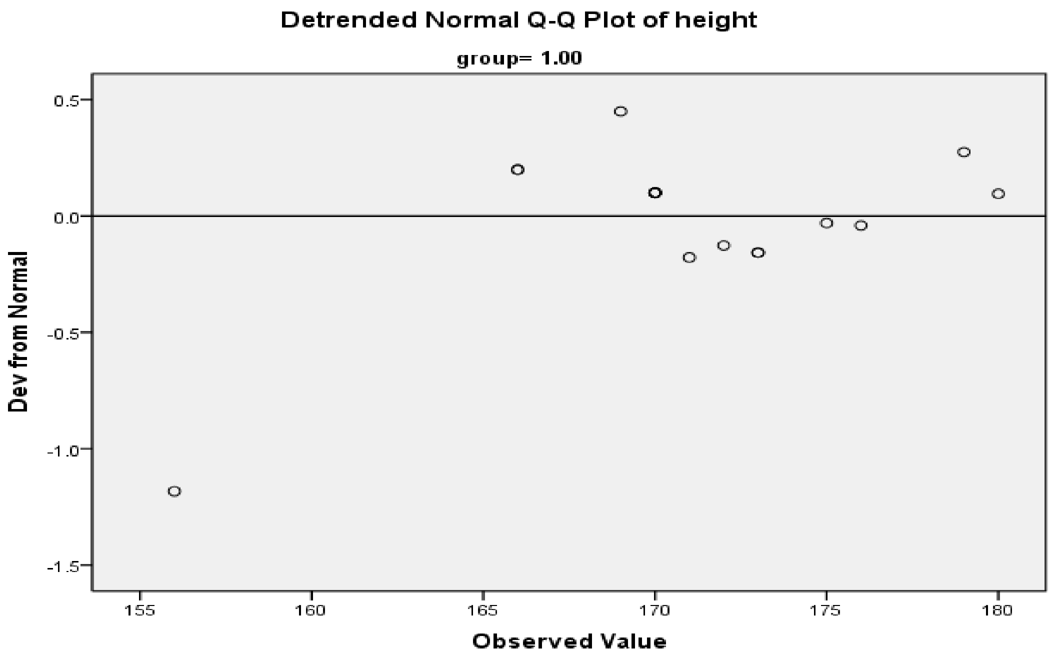

Detrended Normal Q-Q Plot of weight
group= 1.00
Dev from Normal
Observed Value
1.5
1.0
0.5
0.0
-0.5
-1.0
60
70
80
90
100
110
120

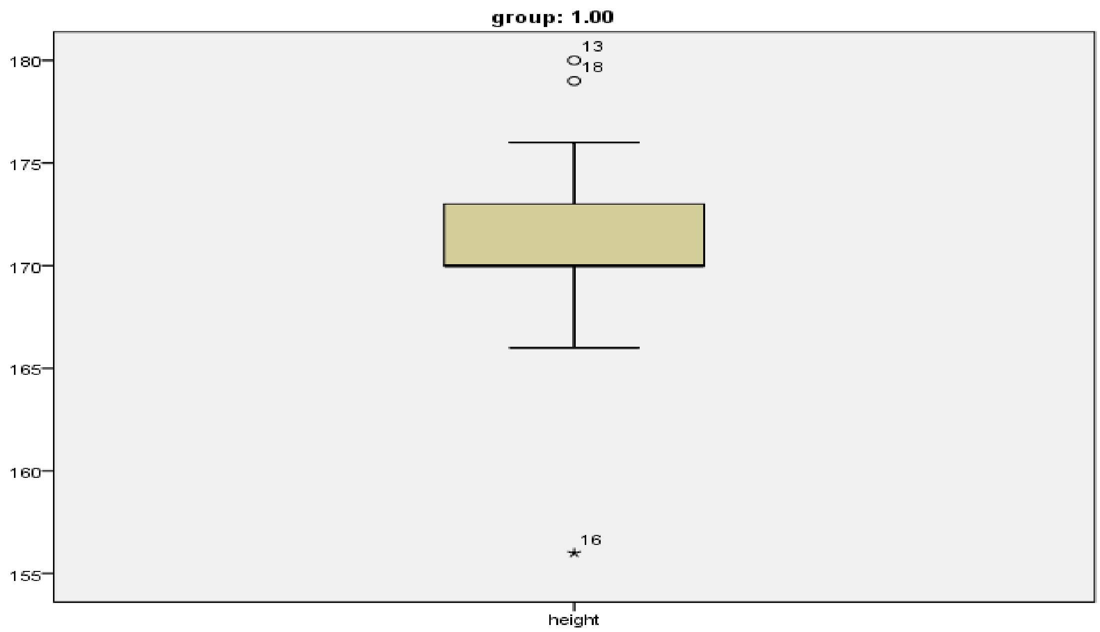

group: 1.00
120
110
100
90
80
70
60
18
13
6
weight

Height


Histogram
group= 1.00
Mean = 170.89
Std. Dev. = 5.312
N = 18
Frequency
6
4
2
0
155.00
160.00
165.00
170.00
175.00
180.00
height


Normal Q-Q Plot of height
group= 1.00
Expected Normal
Observed Value
2
1
0
-1
-2
-3
-4
150
160
170
180


Detrended Normal Q-Q Plot of height
group= 1.00
Dev from Normal
0.5
0.0
-0.5
-1.0
-1.5
155
160
165
170
175
180
Observed Value


group: 1.00
180
175
170
165
160
155
13
18
16
height

Daily tasks group = 2.00

Tests of Normality[b]

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
Age	.085	53	.200*	.956	53	.048
Weight	.139	53	.012	.933	53	.005
Height	.167	53	.001	.948	53	.021

a. Lilliefors Significance Correction

*. This is a lower bound of the true significance.

b. group = 2.00

Case Processing Summary[a]

	Cases					
	Valid		Missing		Total	
	N	Percent	N	Percent	N	Percent
Age	53	100.0%	0	.0%	53	100.0%
Weight	53	100.0%	0	.0%	53	100.0%
Height	53	100.0%	0	.0%	53	100.0%

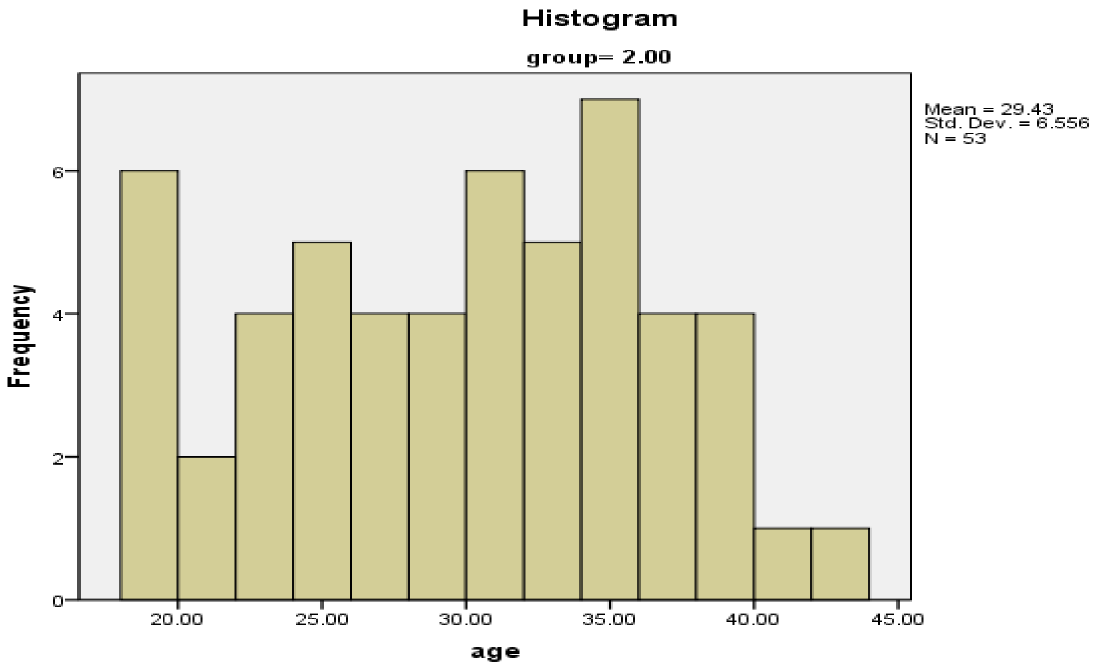
a. group = 2.00

Descriptives[a]

			Statistic	Std. Error
age	Mean		29.4340	.90054
	95% Confidence Interval for Mean	Lower Bound	27.6269	
		Upper Bound	31.2410	
	5% Trimmed Mean		29.3983	
	Median		30.0000	
	Variance		42.981	
	Std. Deviation		6.55600	
	Minimum		19.00	
	Maximum		42.00	
	Range		23.00	
	Interquartile Range		11.00	
	Skewness		-.095	.327
	Kurtosis		-1.064	.644
weight	Mean		73.4528	1.46823
	95% Confidence Interval for Mean	Lower Bound	70.5066	
		Upper Bound	76.3991	
	5% Trimmed Mean		72.9748	
	Median		71.0000	
	Variance		114.253	
	Std. Deviation		10.68890	
	Minimum		50.00	
	Maximum		107.00	
	Range		57.00	
	Interquartile Range		12.00	
	Skewness		.872	.327
	Kurtosis		1.659	.644
height	Mean		171.3208	.73812
	95% Confidence Interval for Mean	Lower Bound	169.8396	
		Upper Bound	172.8019	
	5% Trimmed Mean		171.3994	
	Median		171.0000	
	Variance		28.876	
	Std. Deviation		5.37363	
	Minimum		156.00	
	Maximum		186.00	
	Range		28.00	
	Interquartile Range		4.50	
	Skewness		-.029	.327
	Kurtosis		1.281	.644

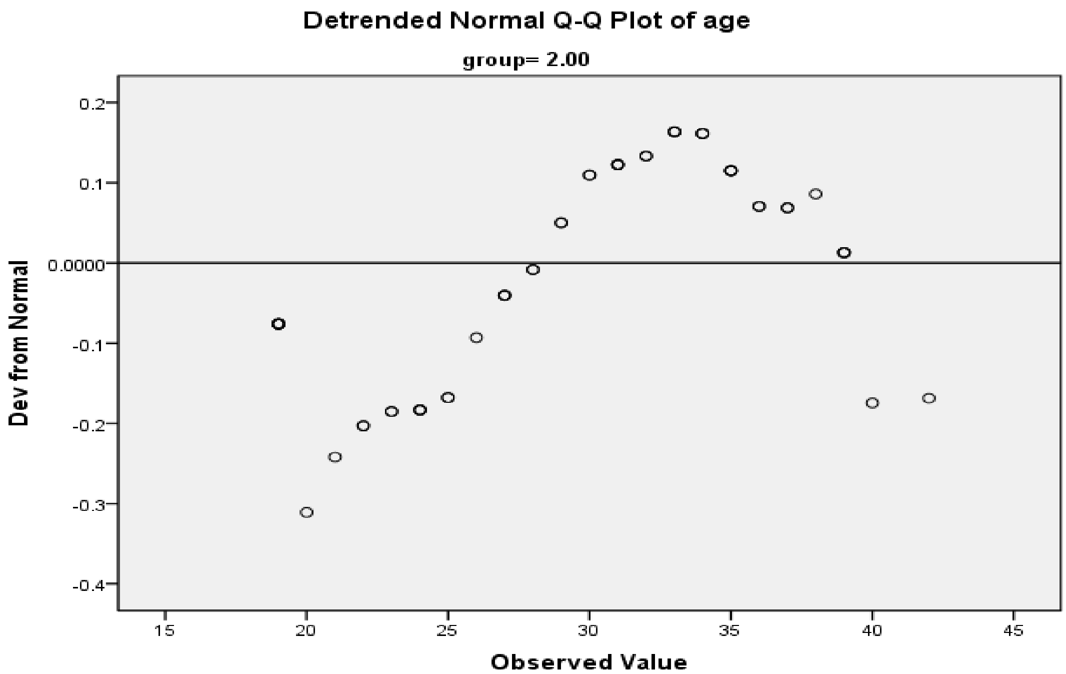
a. group = 2.00

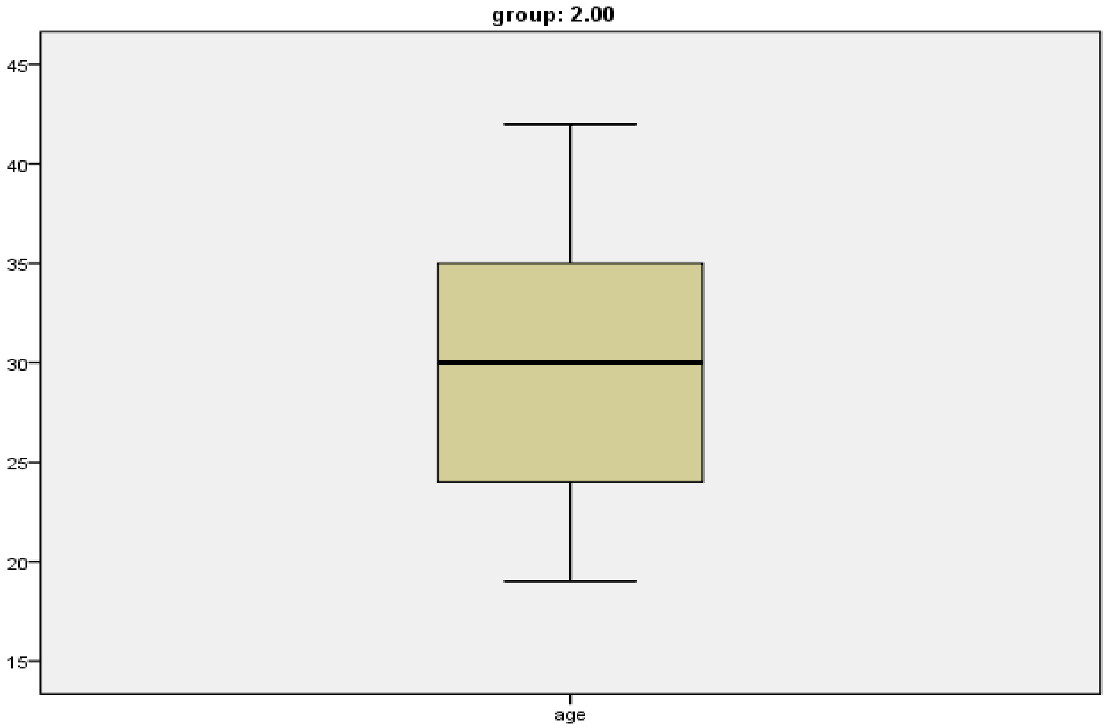
Age



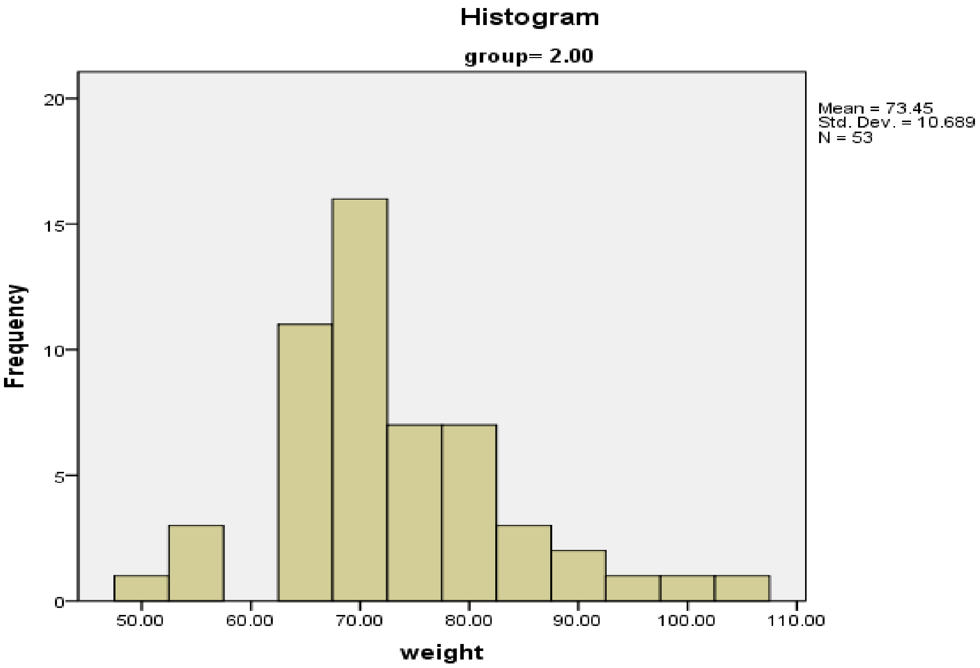
Histogram
group= 2.00
Mean = 29.43
Std. Dev. = 6.556
N = 53
Frequency
0
2
4
6
20.00
25.00
30.00
35.00
40.00
45.00
age

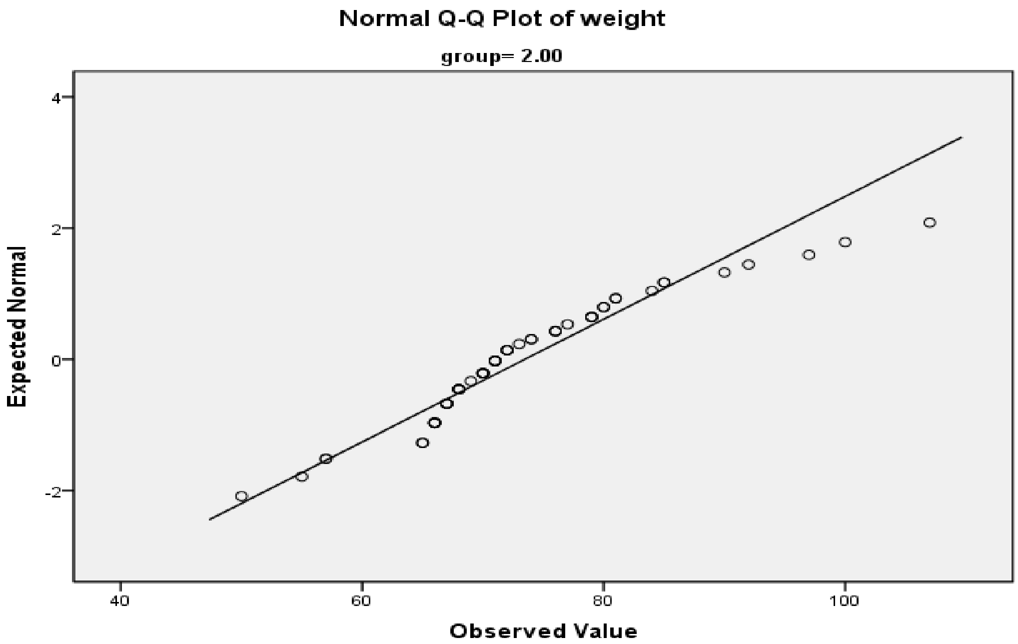

Normal Q-Q Plot of age
group= 2.00
Expected Normal
Observed Value
3
2
1
0
-1
-2
15
20
25
30
35
40
45

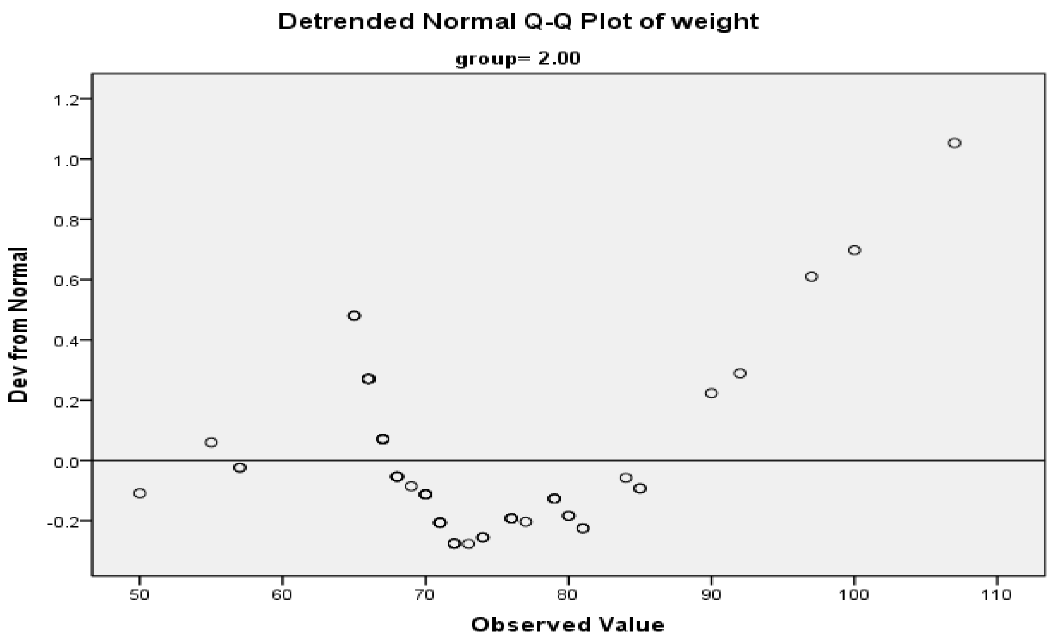

Detrended Normal Q-Q Plot of age
group= 2.00
Dev from Normal
Observed Value
0.2
0.1
0.0000
-0.1
-0.2
-0.3
-0.4
15
20
25
30
35
40
45

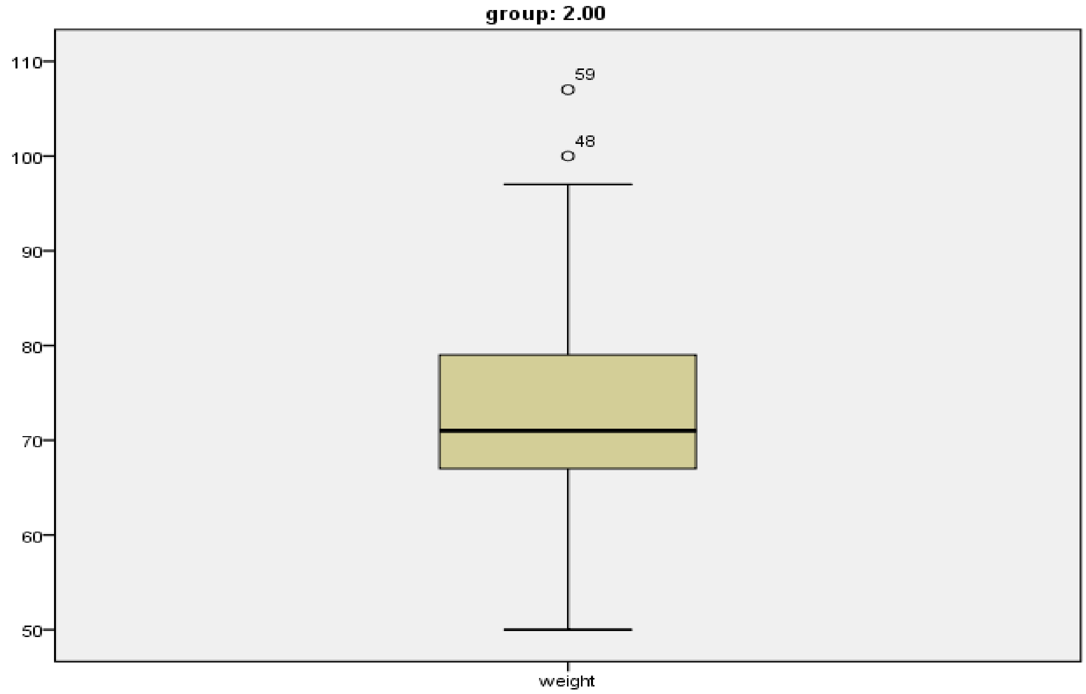

group: 2.00
45
40
35
30
25
20
15
age

Weight

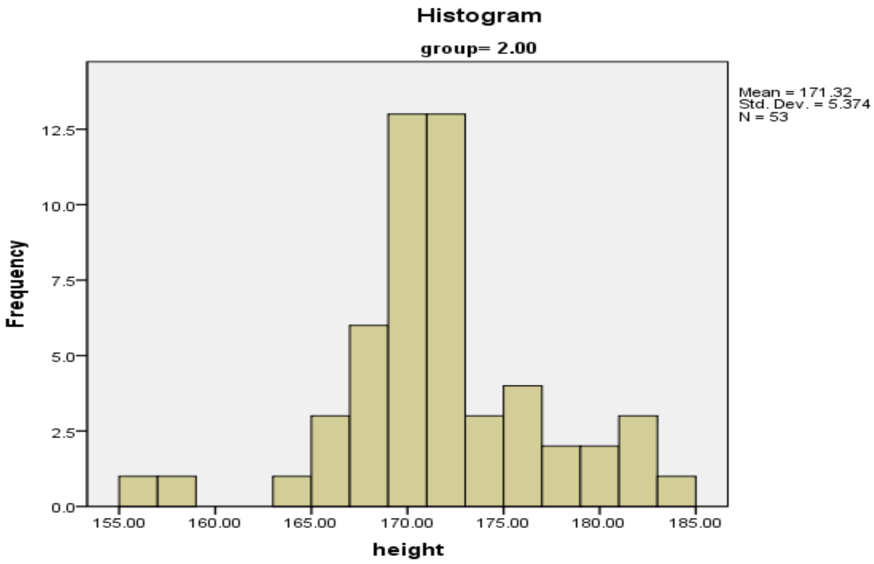

Histogram
group= 2.00
Mean = 73.45
Std. Dev. = 10.689
N = 53
Frequency
20
15
10
5
0
50.00
60.00
70.00
80.00
90.00
100.00
110.00
weight

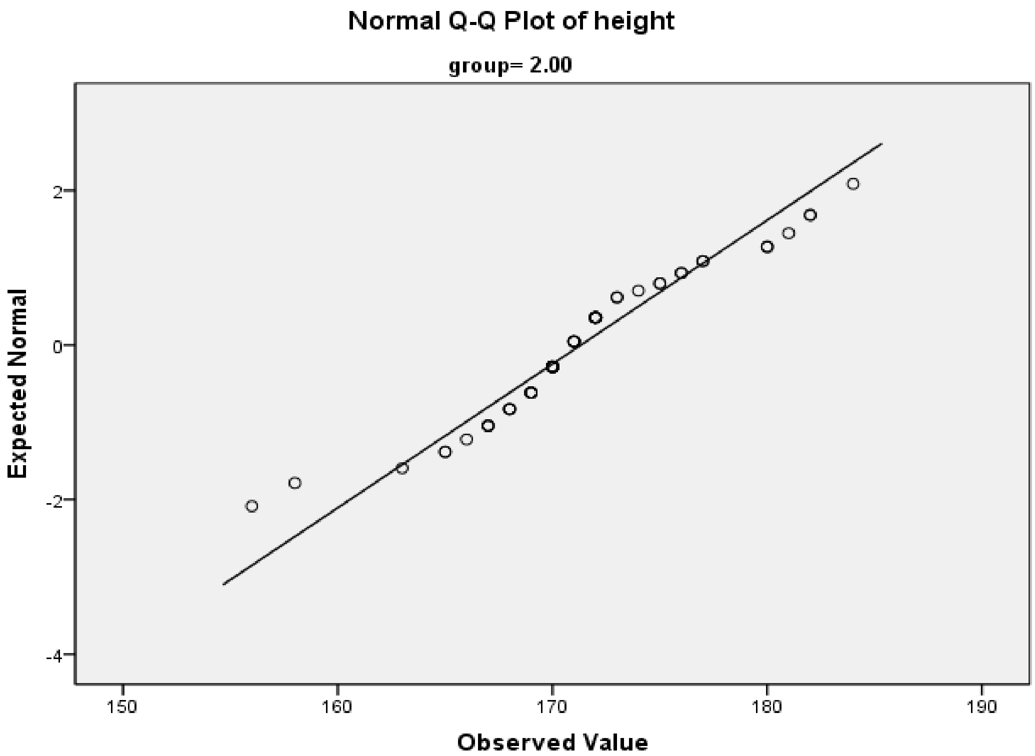

Normal Q-Q Plot of weight
group= 2.00
Expected Normal
Observed Value
4
2
0
-2
40
60
80
100

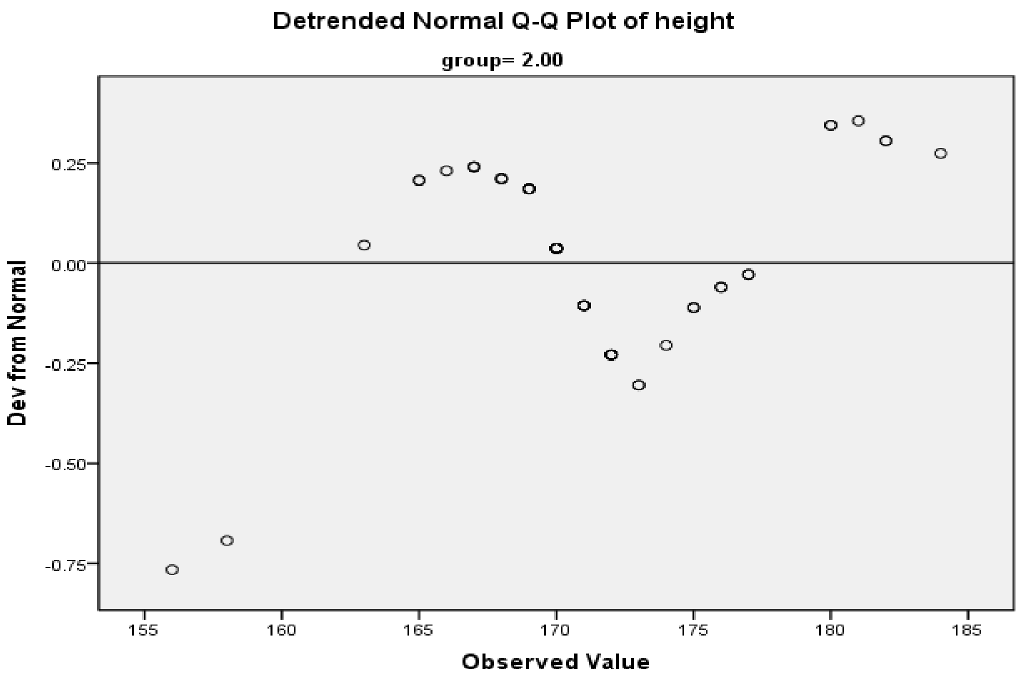

Detrended Normal Q-Q Plot of weight
group= 2.00
Dev from Normal
Observed Value
1.2
1.0
0.8
0.6
0.4
0.2
0.0
-0.2
50
60
70
80
90
100
110

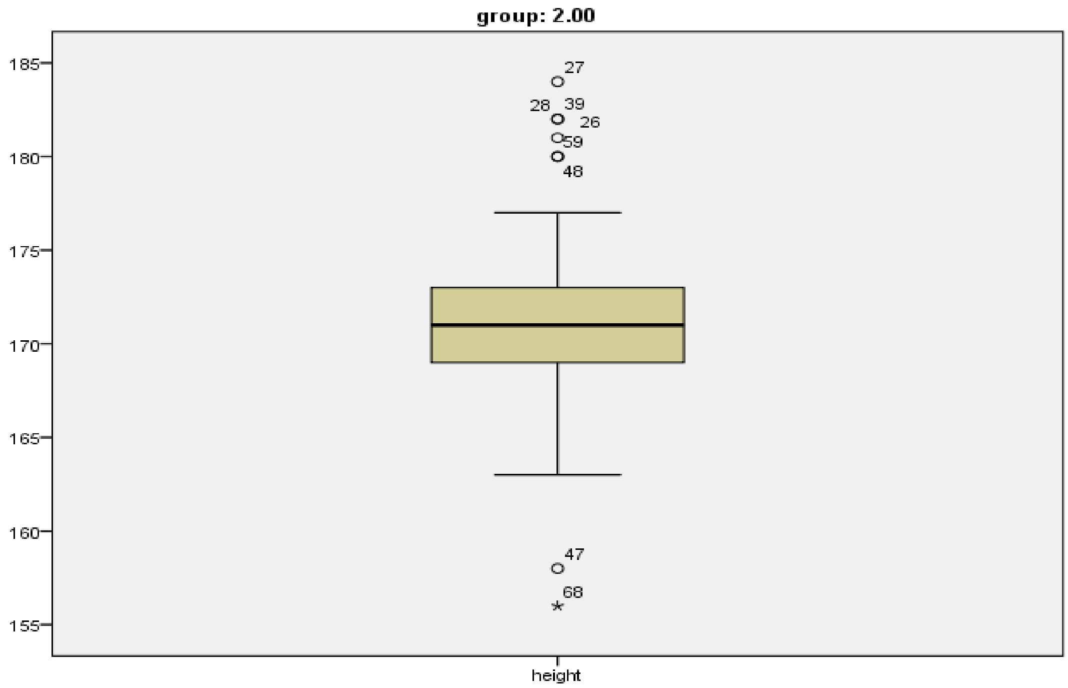


Height


Histogram
group= 2.00
Mean = 171.32
Std. Dev. = 5.374
N = 53
Frequency
0.0
2.5
5.0
7.5
10.0
12.5
155.00
160.00
165.00
170.00
175.00
180.00
185.00
height


Normal Q-Q Plot of height
group= 2.00
Expected Normal
2
0
-2
-4
150
160
170
180
190
Observed Value


Detrended Normal Q-Q Plot of height
group= 2.00
Dev from Normal
0.25
0.00
-0.25
-0.50
-0.75
155
160
165
170
175
180
185
Observed Value


group: 2.00
185
180
175
170
165
160
155
27
28
39
26
59
48
47
68
height

ii. Flexion task

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
ULx ROM	.111	53	.098	.968	53	.159
LLx ROM	.132	53	.021	.955	53	.046
WLx ROM	.156	53	.003	.959	53	.069

a. Lilliefors Significance Correction

Case Processing Summary

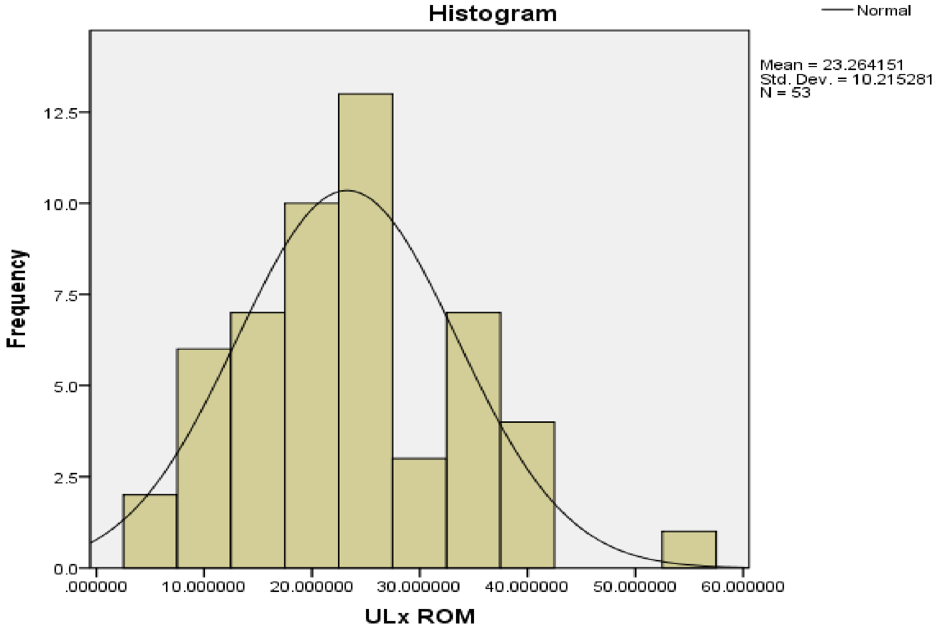
	Cases					
	Valid		Missing		Total	
	N	Percent	N	Percent	N	Percent
ULx ROM	53	100.0%	0	.0%	53	100.0%
LLx ROM	53	100.0%	0	.0%	53	100.0%
WLx ROM	53	100.0%	0	.0%	53	100.0%

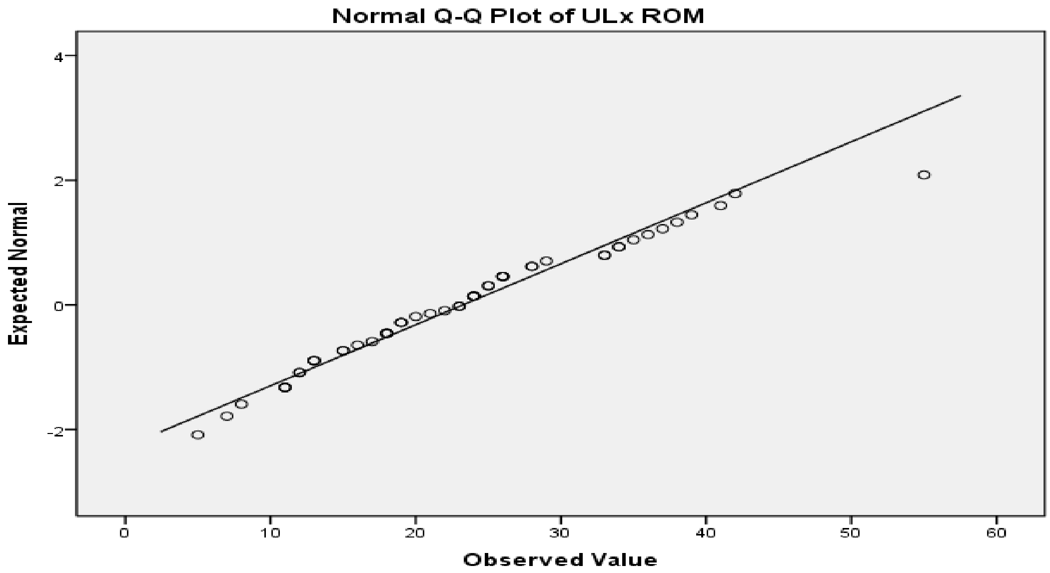
Descriptives

			Statistic	Std. Error
ULx ROM	Mean		23.26415094	1.403176715
	95% Confidence Interval for Mean	Lower Bound	20.44847087	
		Upper Bound	26.07983102	
	5% Trimmed Mean		22.89622642	
	Median		23.00000000	
	Variance		104.352	
	Std. Deviation		10.215280680	
	Minimum		5.000000	
	Maximum		55.000000	
	Range		50.000000	
	Interquartile Range		13.000000	
	Skewness		.639	.327
	Kurtosis		.505	.644
LLx ROM	Mean		36.03773585	1.843468497
	95% Confidence Interval for Mean	Lower Bound	32.33854568	
		Upper Bound	39.73692602	
	5% Trimmed Mean		36.59224319	
	Median		38.00000000	
	Variance		180.114	
	Std. Deviation		13.420653234	
	Minimum		-9.000000	
	Maximum		63.000000	
	Range		72.000000	
	Interquartile Range		16.500000	
	Skewness		-.859	.327
	Kurtosis		1.360	.644
WLx ROM	Mean		59.30188679	1.968287200
	95% Confidence Interval for Mean	Lower Bound	55.35222957	
		Upper Bound	63.25154401	
	5% Trimmed Mean		59.71069182	
	Median		60.00000000	
	Variance		205.330	
	Std. Deviation		14.329347113	
	Minimum		22.000000	
	Maximum		92.000000	
	Range		70.000000	

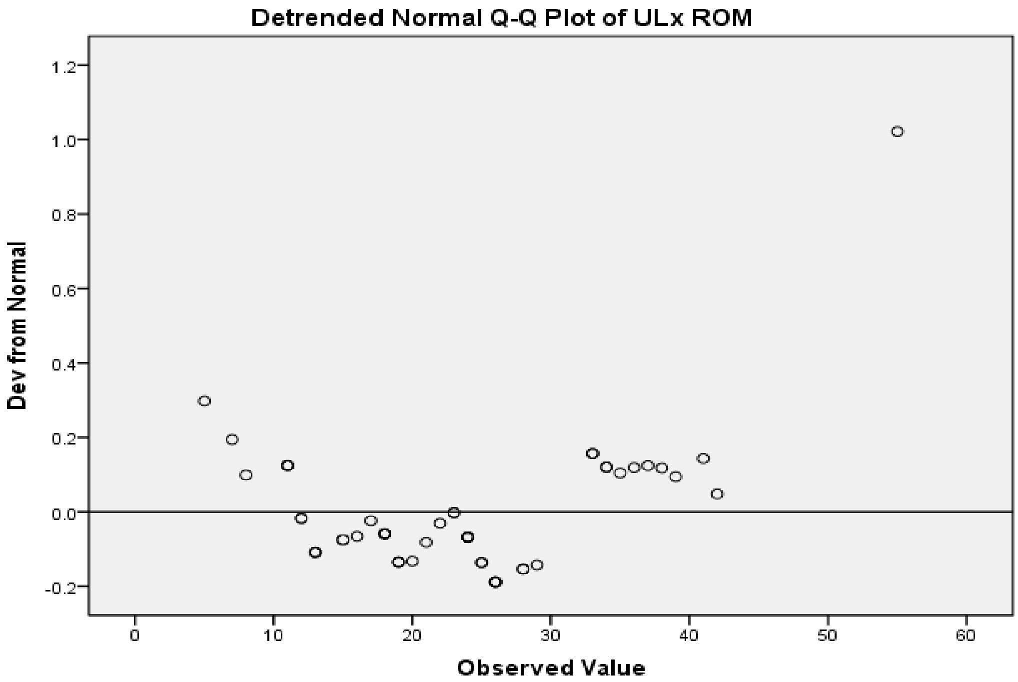
	Interquartile Range	13.000000	
	Skewness	-.522	.327
	Kurtosis	.771	.644

ULx ROM

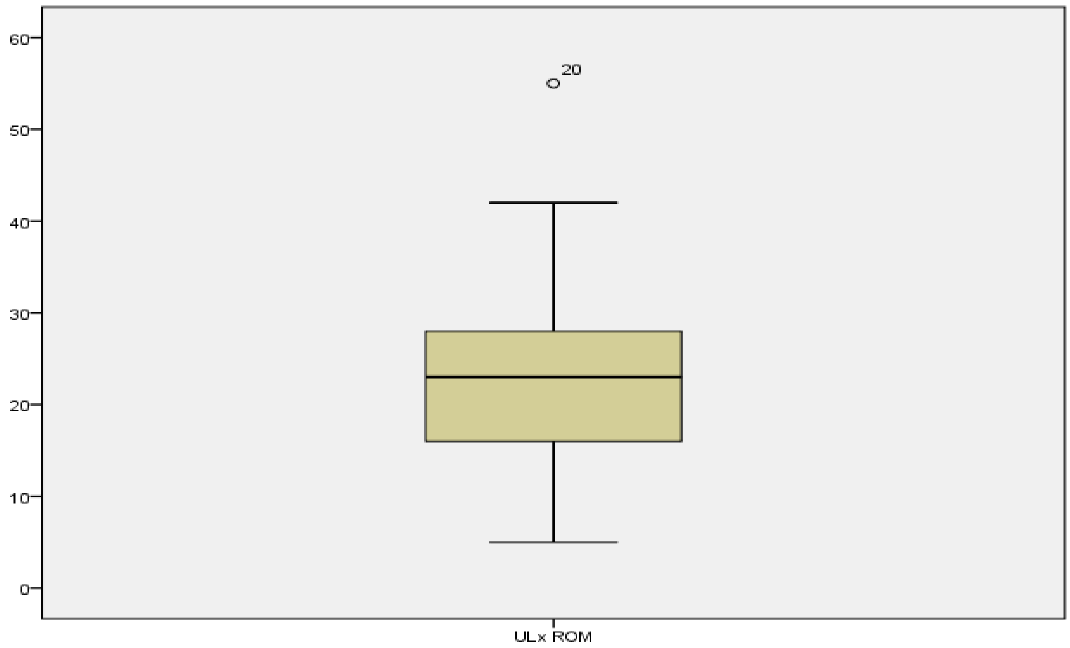




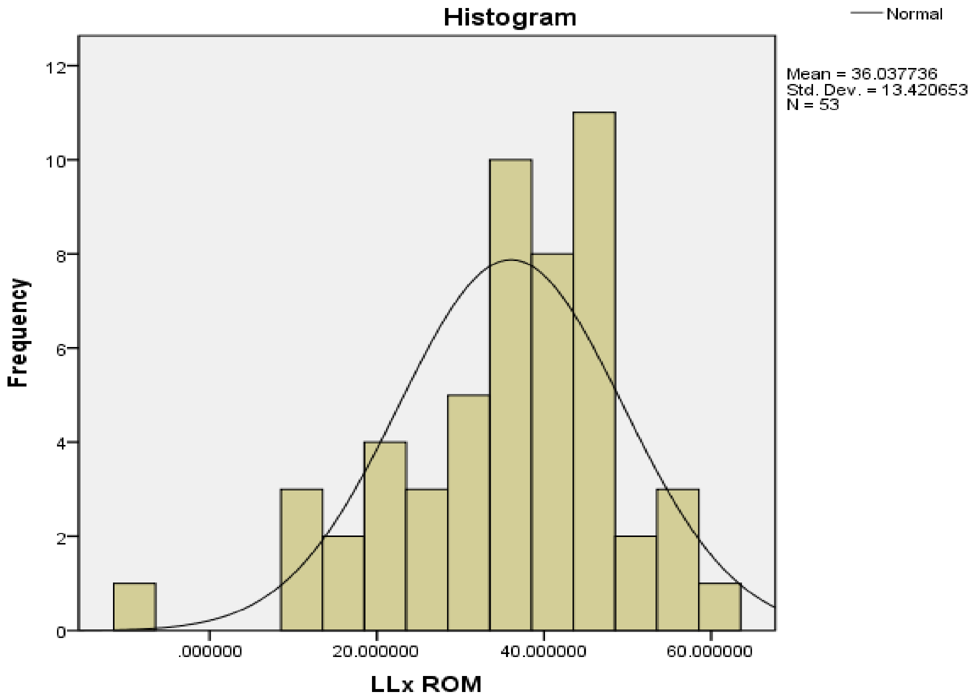
Normal Q-Q Plot of ULx ROM
Expected Normal
Observed Value
4
2
0
-2
0
10
20
30
40
50
60

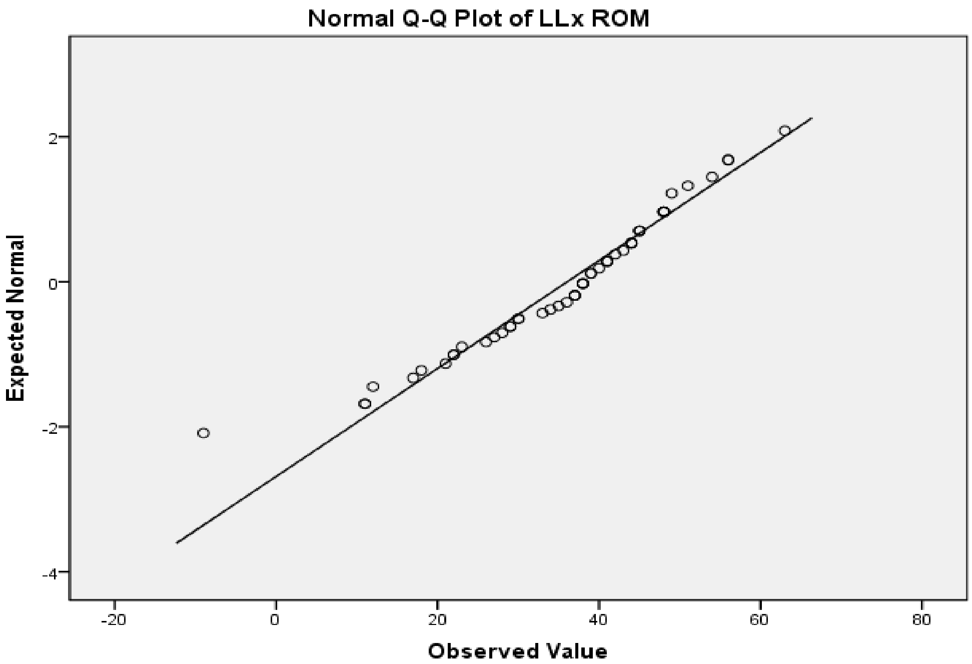


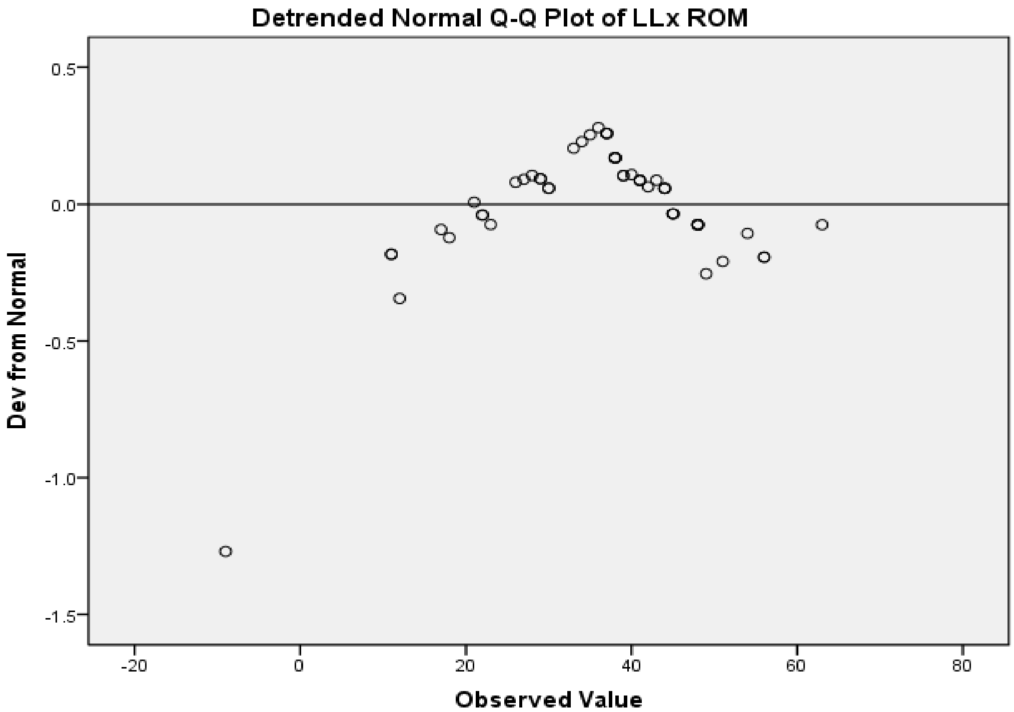
Detrended Normal Q-Q Plot of ULx ROM
Dev from Normal
Observed Value
1.2
1.0
0.8
0.6
0.4
0.2
0.0
-0.2
0
10
20
30
40
50
60

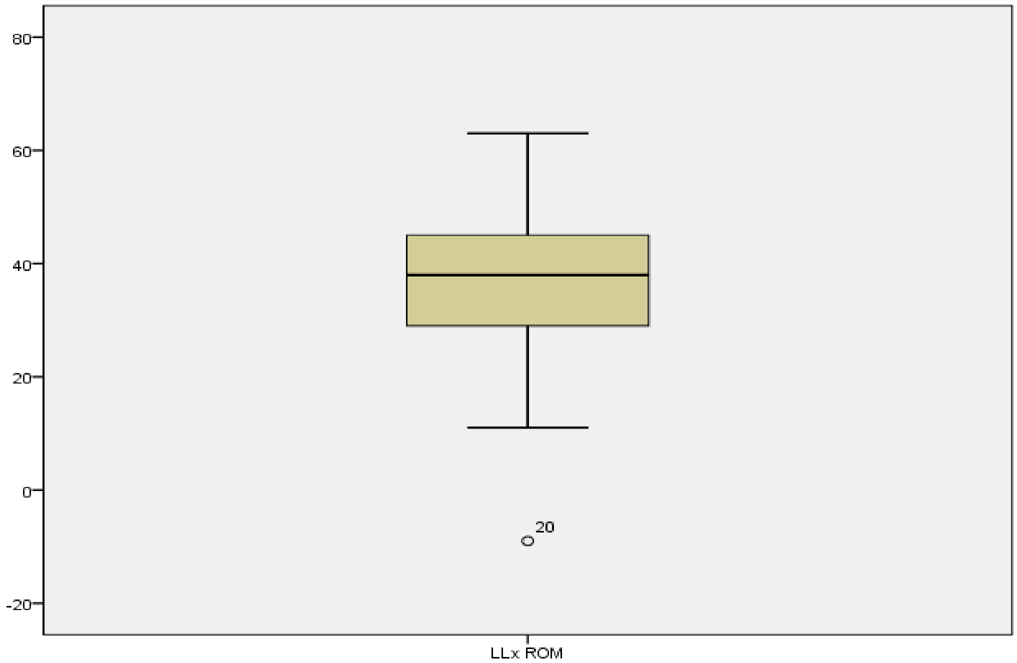


LLx ROM

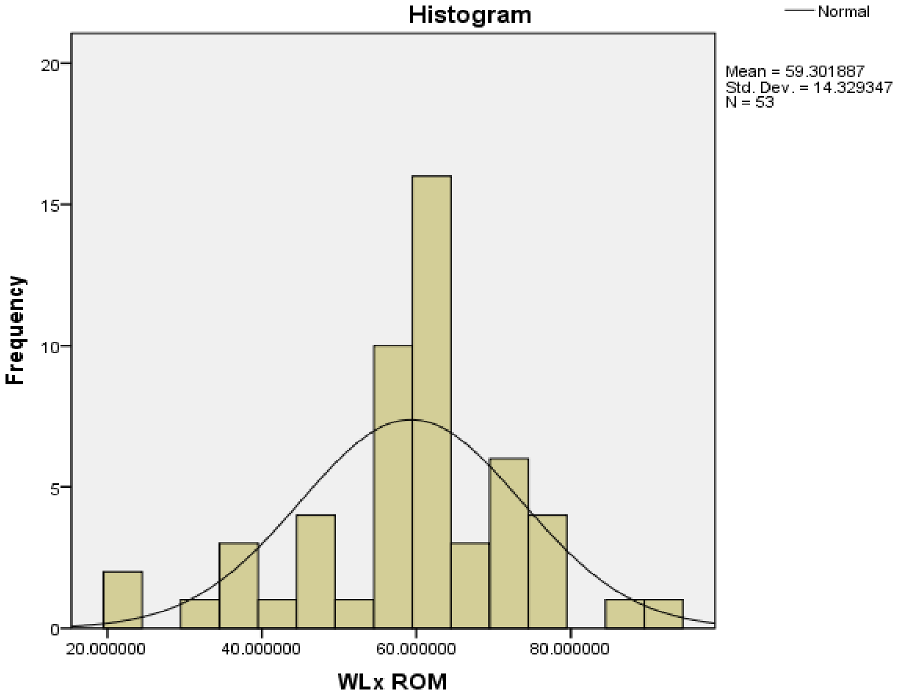

Histogram
Normal
Mean = 36.037736
Std. Dev. = 13.420653
N = 53
Frequency
0
2
4
6
8
10
12
.000000
20.000000
40.000000
60.000000
LLx ROM


Normal Q-Q Plot of LLx ROM
Expected Normal
2
0
-2
-4
-20
0
20
40
60
80
Observed Value

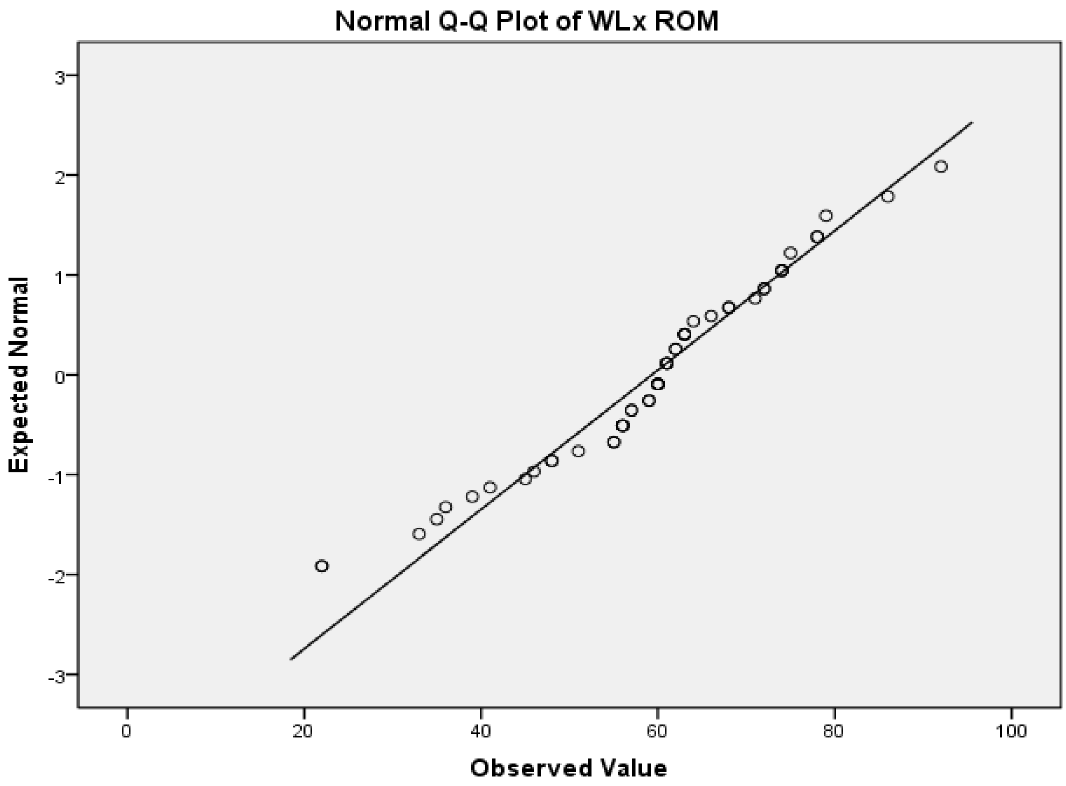

Detrended Normal Q-Q Plot of LLx ROM
0.5
0.0
-0.5
-1.0
-1.5
Dev from Normal
-20
0
20
40
60
80
Observed Value

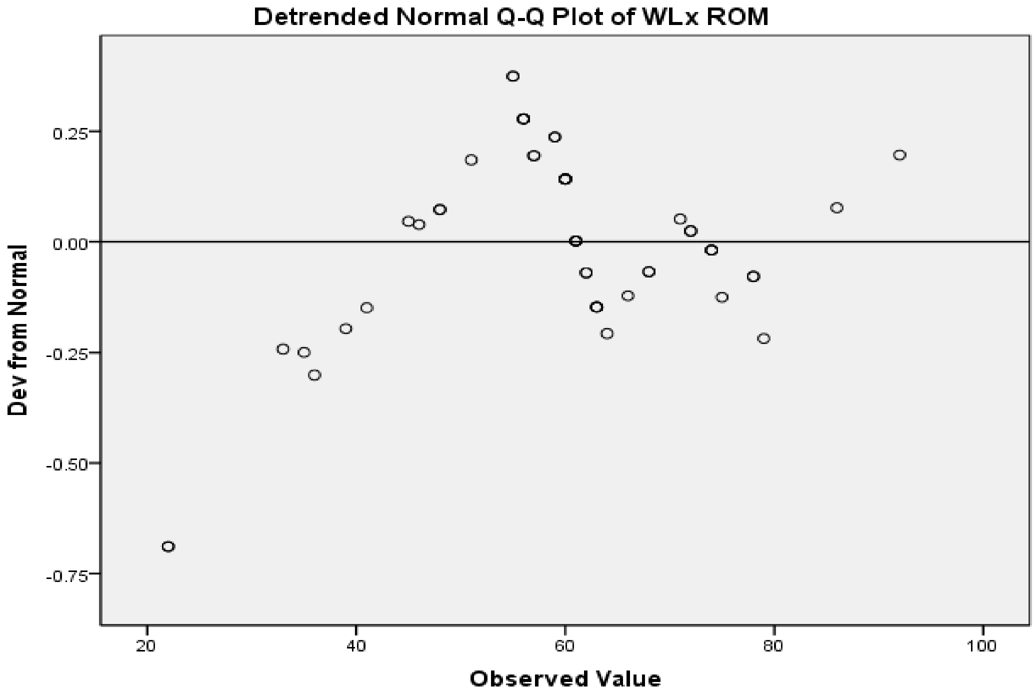

80
60
40
20
0
-20
20
LLx ROM

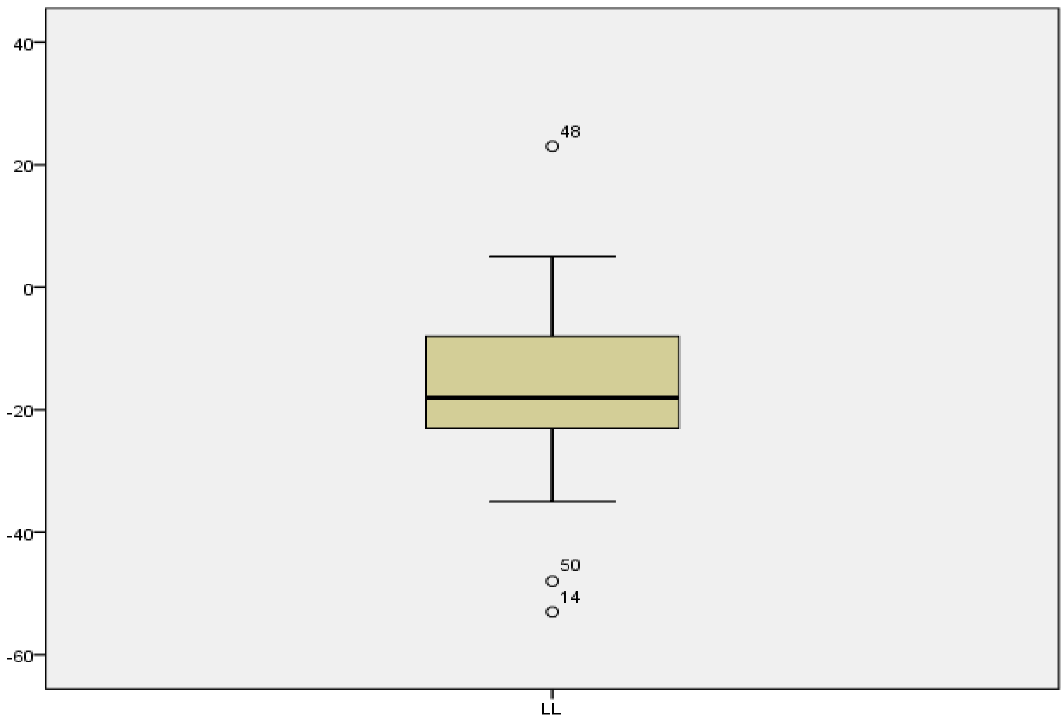
WLx ROM



Histogram
Normal
Mean = 59.301887
Std. Dev. = 14.329347
N = 53
Frequency
20
15
10
5
0
20.000000
40.000000
60.000000
80.000000
WLx ROM


Normal Q-Q Plot of WLx ROM
Expected Normal
Observed Value
3
2
1
0
-1
-2
-3
0
20
40
60
80
100


Detrended Normal Q-Q Plot of WLx ROM
Dev from Normal
Observed Value
0.25
0.00
-0.25
-0.50
-0.75
20
40
60
80
100


40
20
0
-20
-40
-60
48
50
14
LL

iii. Object lifting

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
ULx ROM	.123	53	.043	.953	53	.037
LLx ROM	.113	53	.091	.968	53	.159
WLx ROM	.124	53	.042	.959	53	.064

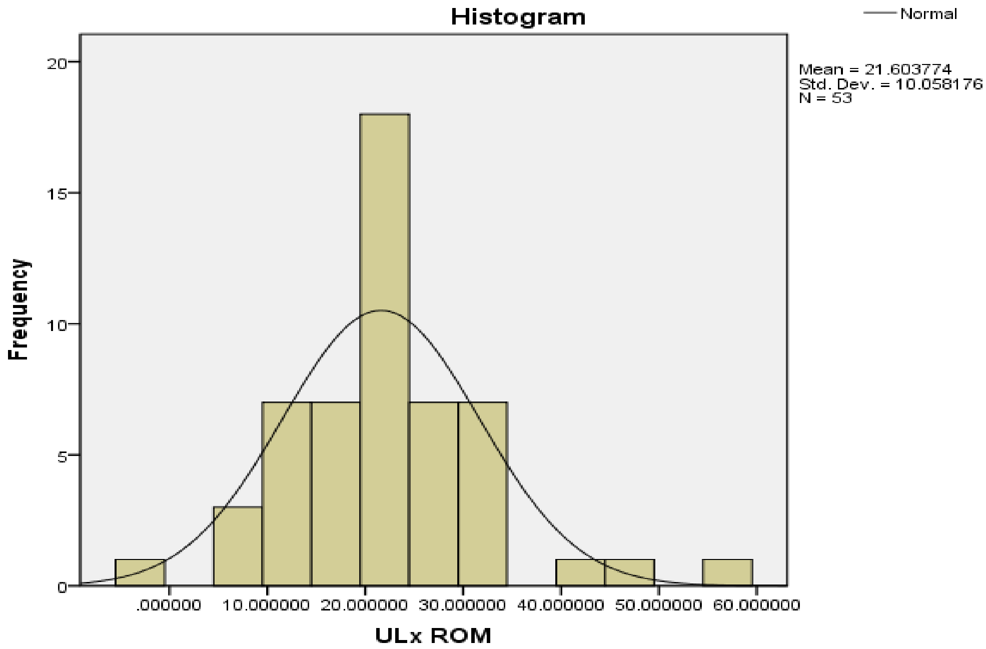
a. Lilliefors Significance Correction

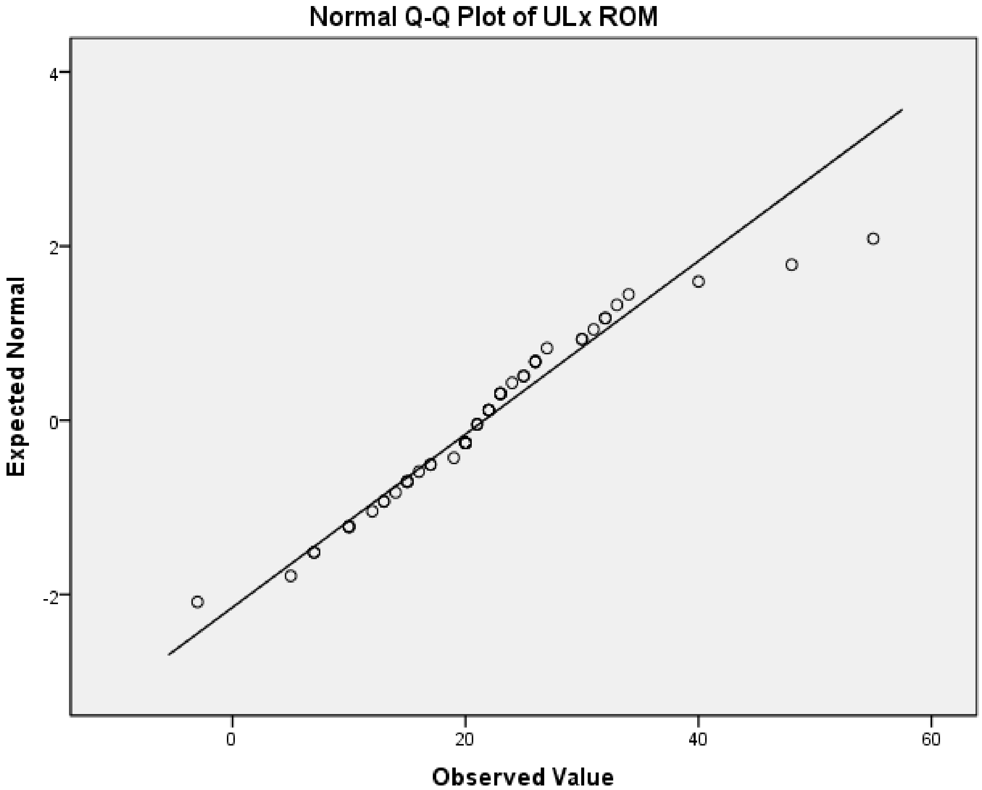
Descriptives

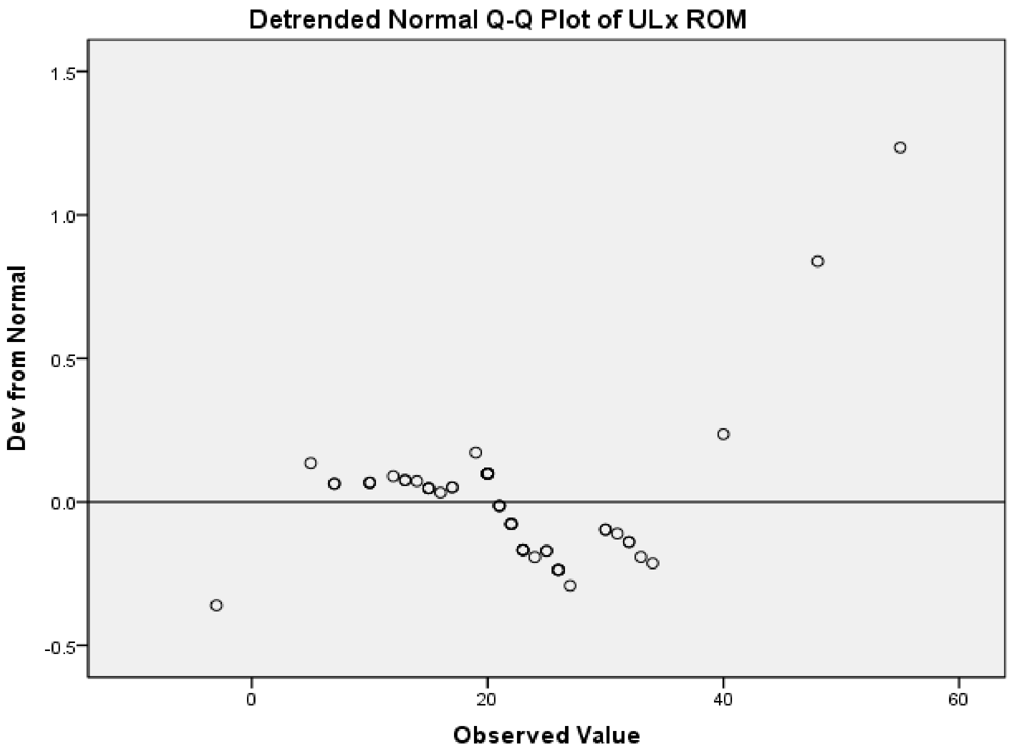
			Statistic	Std. Error
ULx ROM	Mean		21.60377358	1.381596756
	95% Confidence Interval for Mean	Lower Bound	18.83139687	
		Upper Bound	24.37615030	
	5% Trimmed Mean		21.16247379	
	Median		21.00000000	
	Variance		101.167	
	Std. Deviation		10.058176205	
	Minimum		-3.000000	
	Maximum		55.000000	
	Range		58.000000	
	Interquartile Range		11.000000	
	Skewness		.704	.327
	Kurtosis		2.189	.644
LLx ROM	Mean		35.358490566	1.9305981573
	95% Confidence Interval for Mean	Lower Bound	31.484461942	
		Upper Bound	39.232519190	
	5% Trimmed Mean		35.921383648	
	Median		38.000000000	
	Variance		197.542	
	Std. Deviation		14.0549667373	
	Minimum		-10.0000000	
	Maximum		64.0000000	
	Range		74.0000000	
	Interquartile Range		20.0000000	
	Skewness		-.726	.327
	Kurtosis		.926	.644
WLx ROM	Mean		55.96226415	2.208796615
	95% Confidence Interval for Mean	Lower Bound	51.52998948	
		Upper Bound	60.39453882	
	5% Trimmed Mean		56.58385744	
	Median		58.00000000	
	Variance		258.575	
	Std. Deviation		16.080282078	
	Minimum		8.000000	
	Maximum		87.000000	
	Range		79.000000	
	Interquartile Range		21.000000	

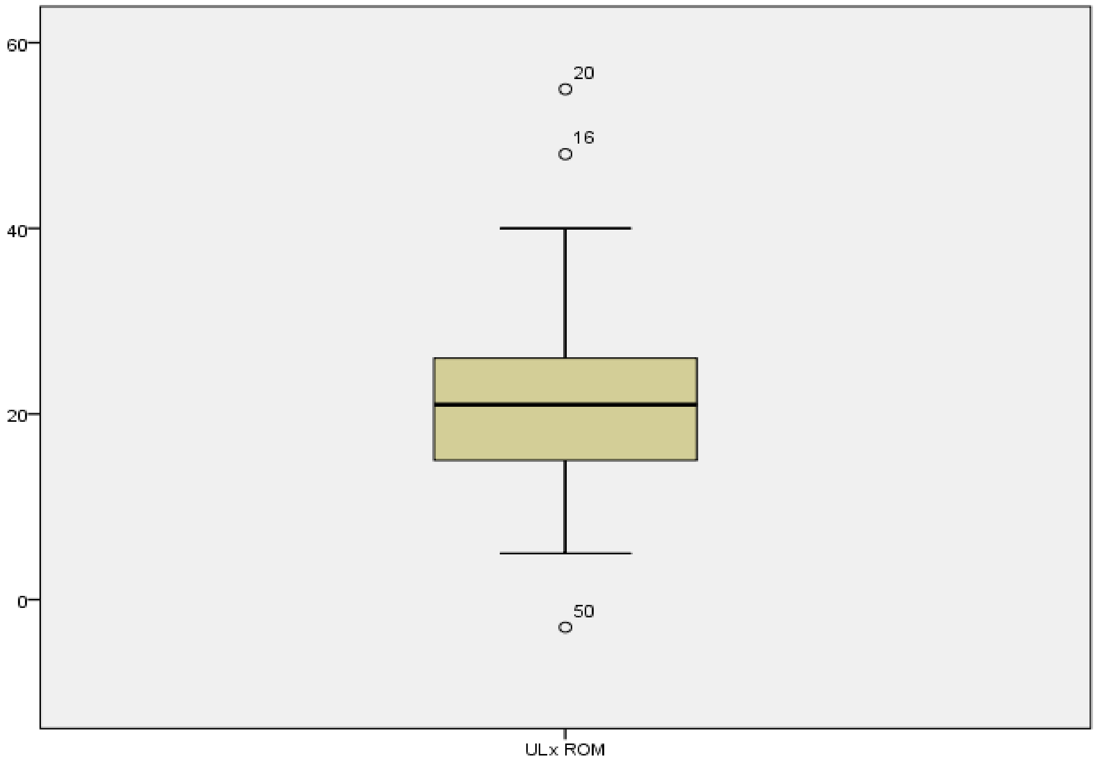
	Skewness	-.759	.327
	Kurtosis	.731	.644

ULx ROM

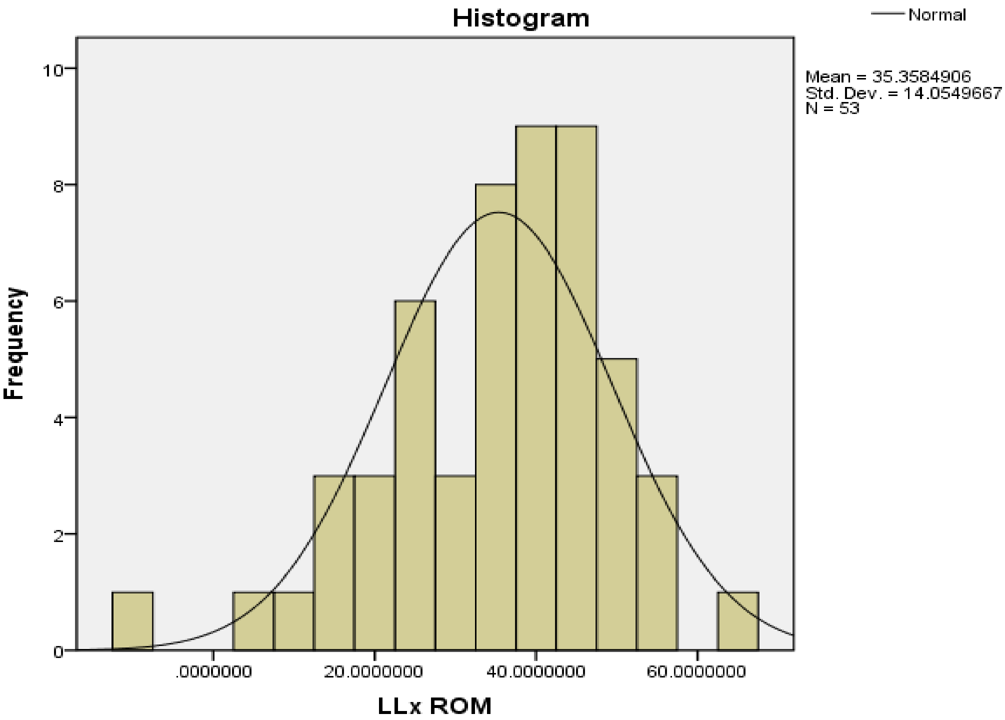



Normal Q-Q Plot of ULx ROM
Expected Normal
Observed Value
4
2
0
-2
0
20
40
60

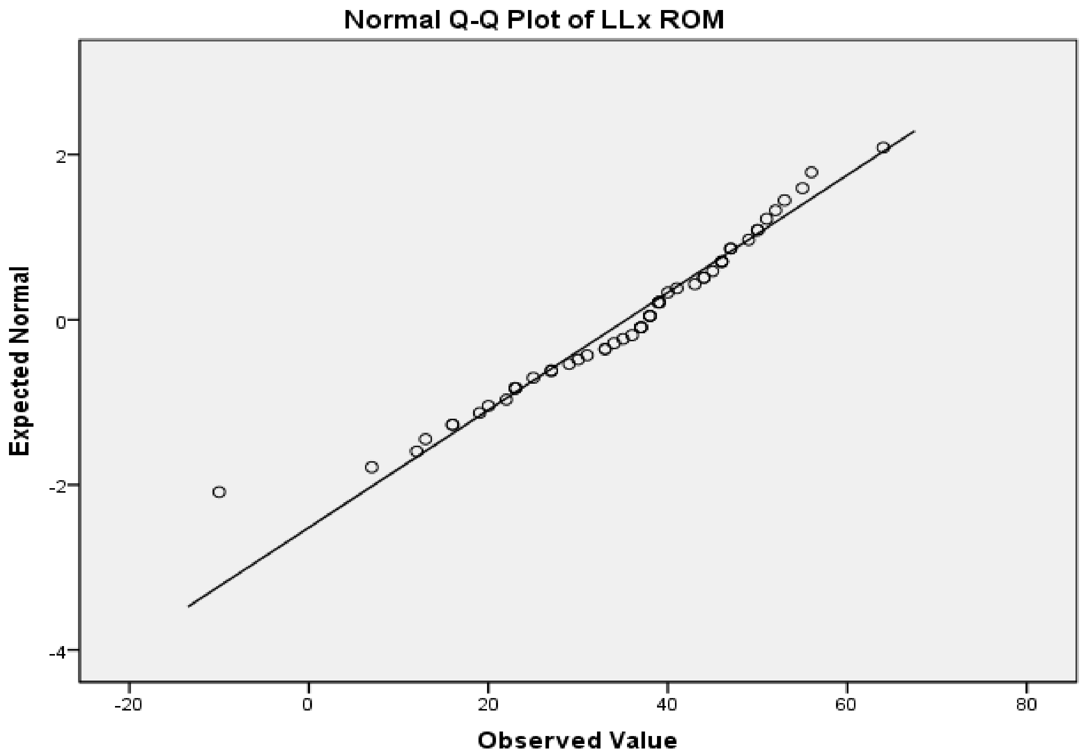

Detrended Normal Q-Q Plot of ULx ROM
Dev from Normal
Observed Value
1.5
1.0
0.5
0.0
-0.5
0
20
40
60

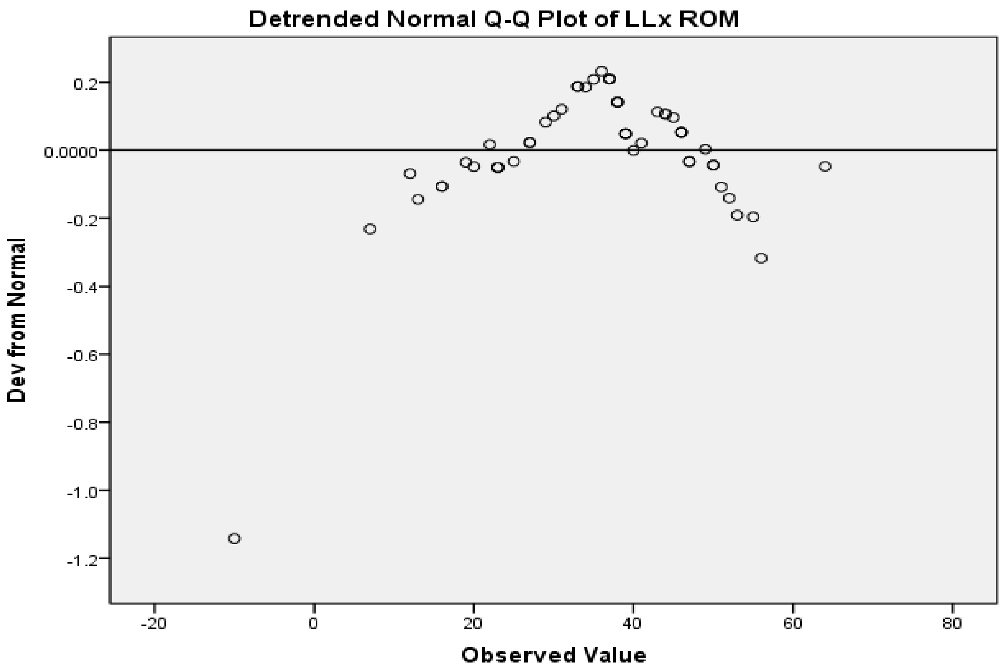

60
40
20
0
20
16
50
ULx ROM

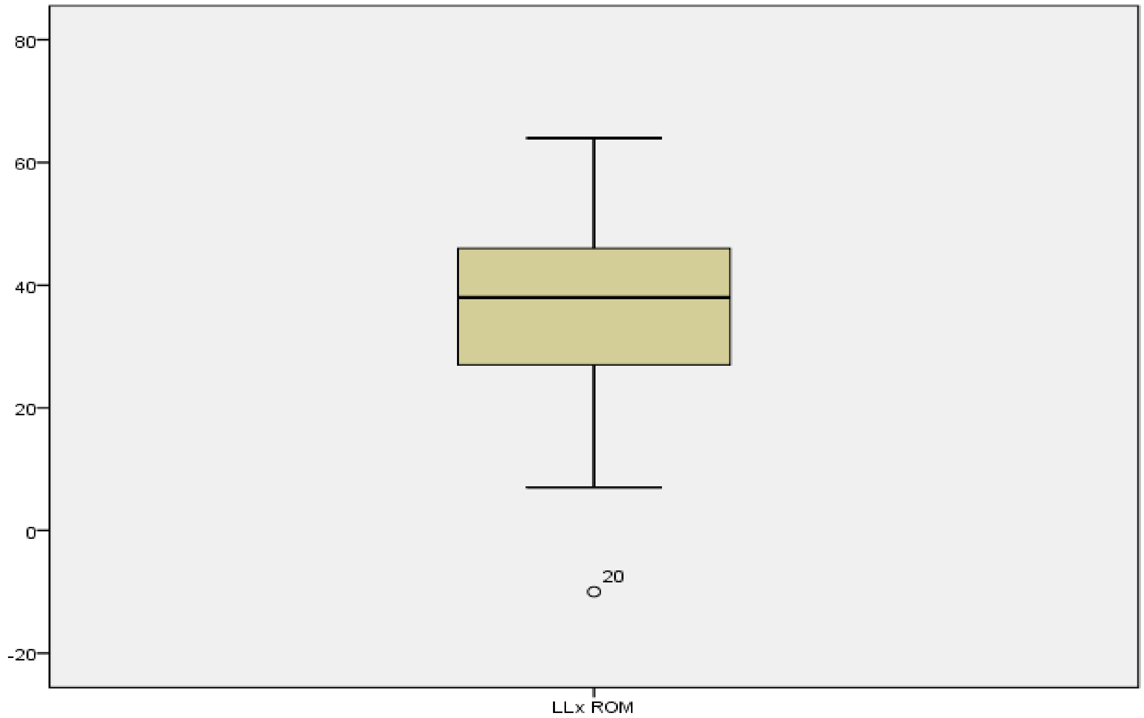
LLx ROM



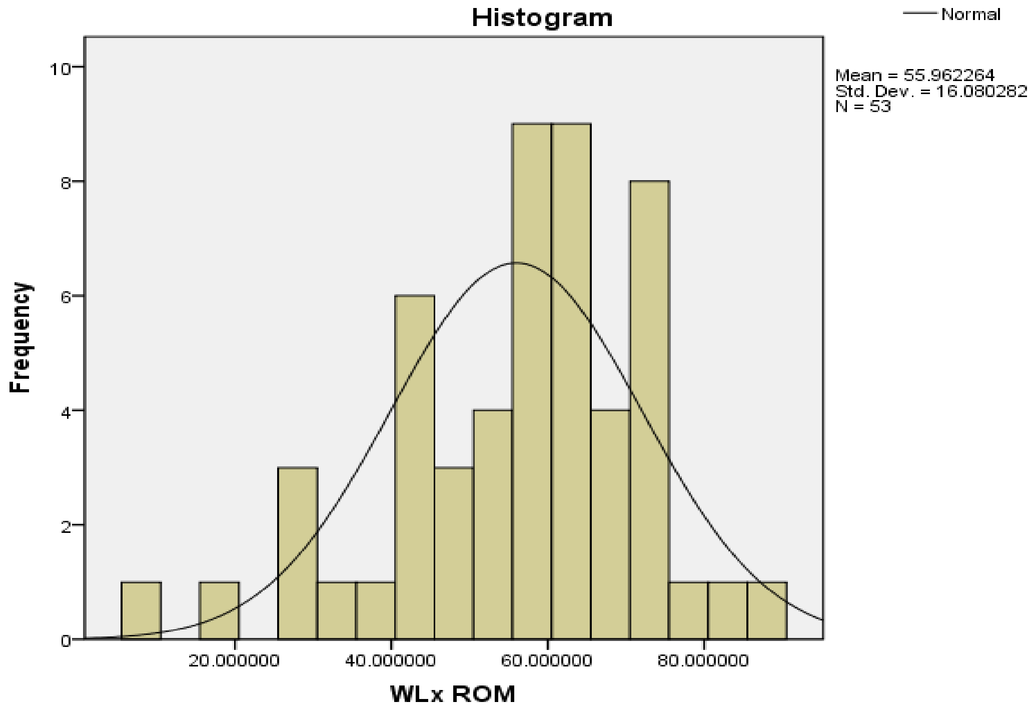
Histogram
Normal
Mean = 35.3584906
Std. Dev. = 14.0549667
N = 53
Frequency
0
2
4
6
8
10
.0000000
20.0000000
40.0000000
60.0000000
LLx ROM

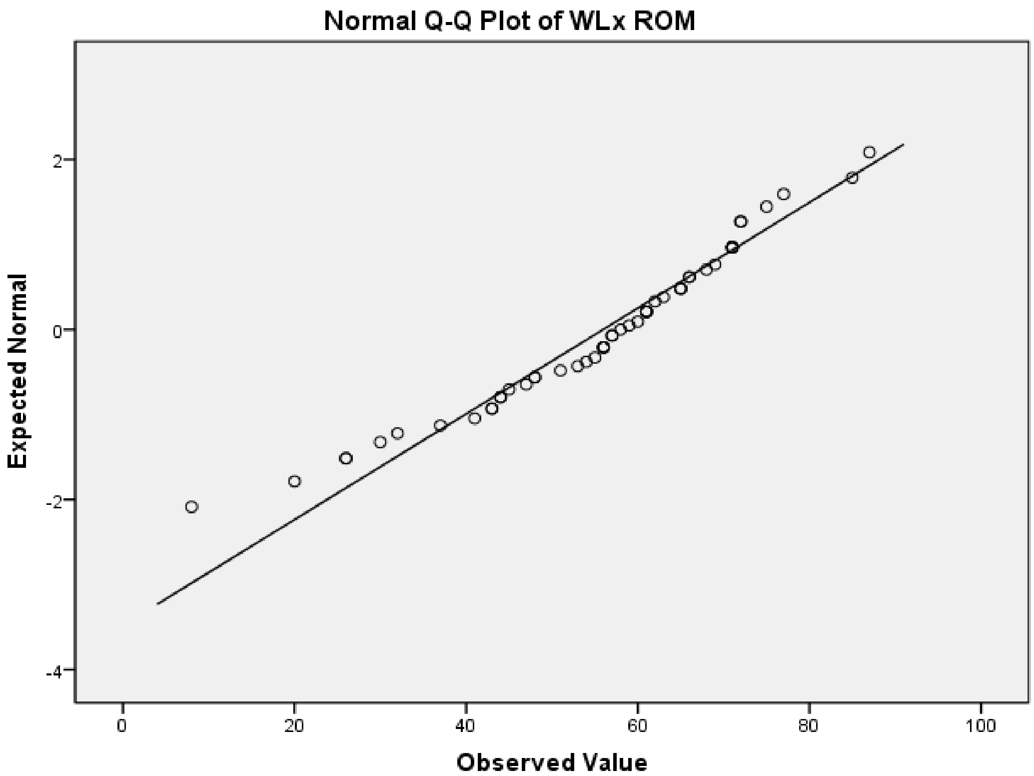

Normal Q-Q Plot of LLx ROM
Expected Normal
2
0
-2
-4
-20
0
20
40
60
80
Observed Value

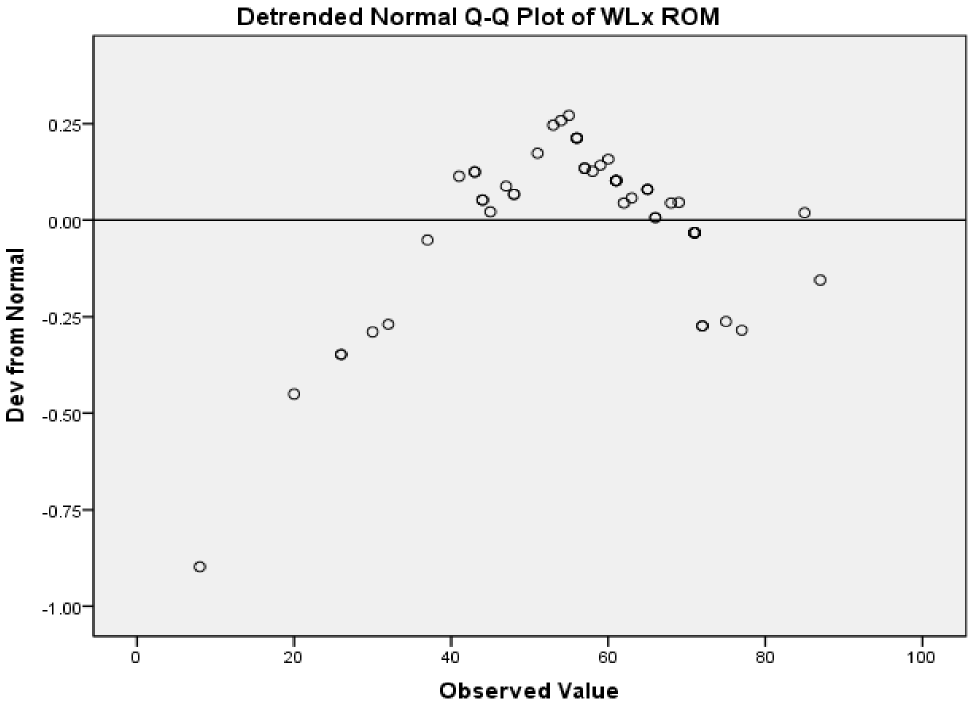

Detrended Normal Q-Q Plot of LLx ROM
Dev from Normal
0.2
0.0000
-0.2
-0.4
-0.6
-0.8
-1.0
-1.2
-20
0
20
40
60
80
Observed Value

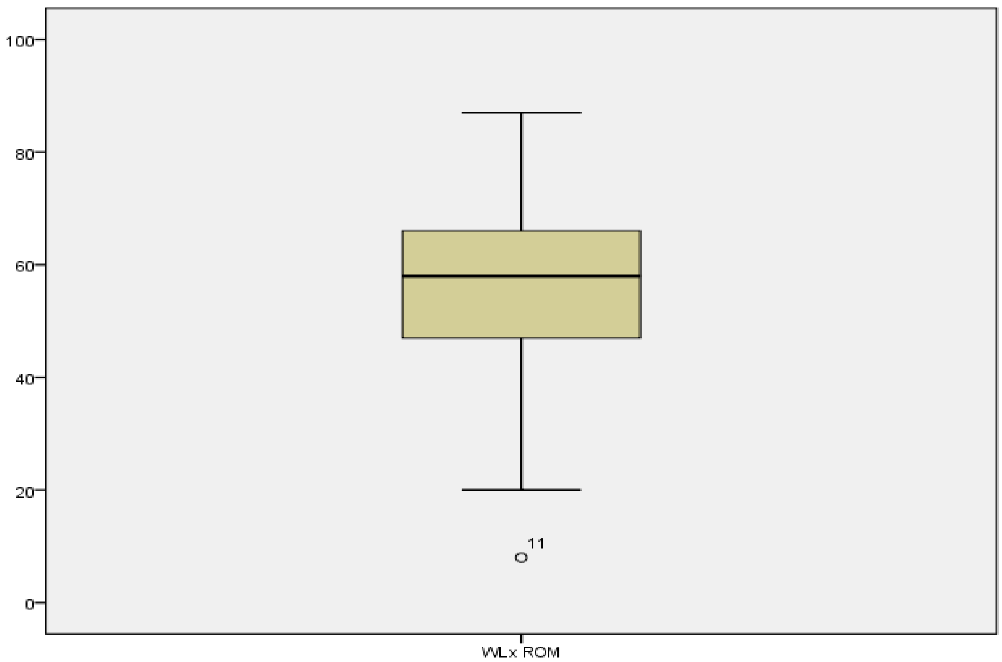


WLx ROM


Histogram
Normal
Mean = 55.962264
Std. Dev. = 16.080282
N = 53
Frequency
0
2
4
6
8
10
20.000000
40.000000
60.000000
80.000000
WLx ROM


Normal Q-Q Plot of WLx ROM
Expected Normal
2
0
-2
-4
0
20
40
60
80
100
Observed Value


Detrended Normal Q-Q Plot of WLx ROM
Dev from Normal
Observed Value
0.25
0.00
-0.25
-0.50
-0.75
-1.00
0
20
40
60
80
100


100
80
60
40
20
0
11
WLx ROM

iv. Standing-Siting

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	Df	Sig.
UL	.085	53	.200*	.970	53	.212
LL	.068	53	.200*	.987	53	.846
WL	.088	53	.200*	.981	53	.542

a. Lilliefors Significance Correction

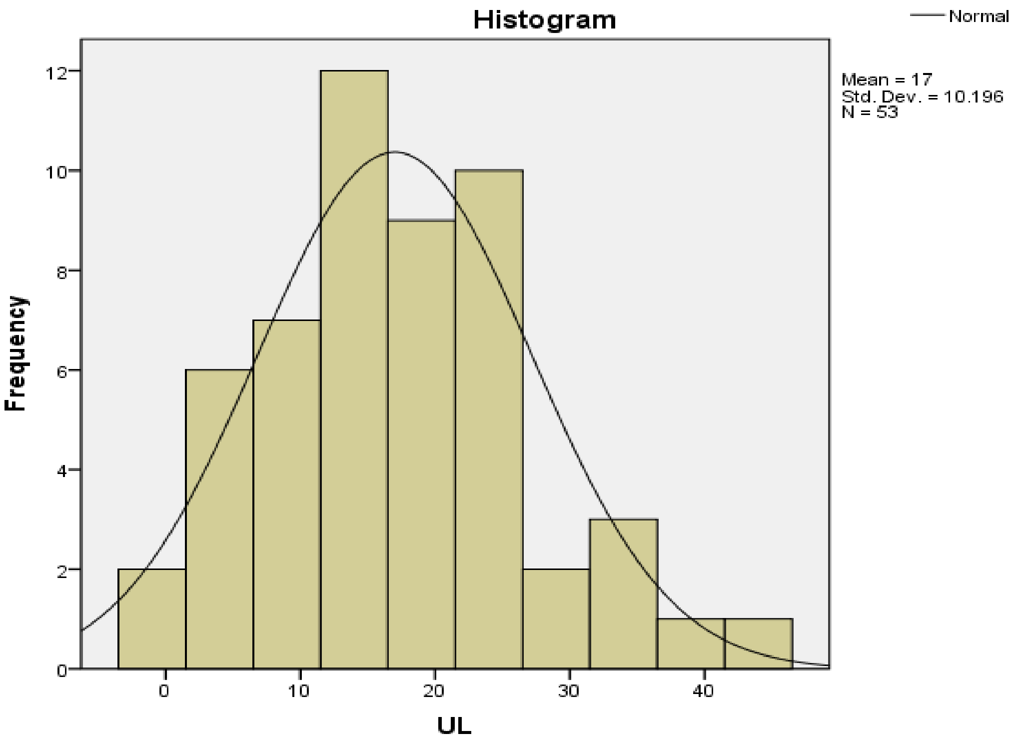
*. This is a lower bound of the true significance.

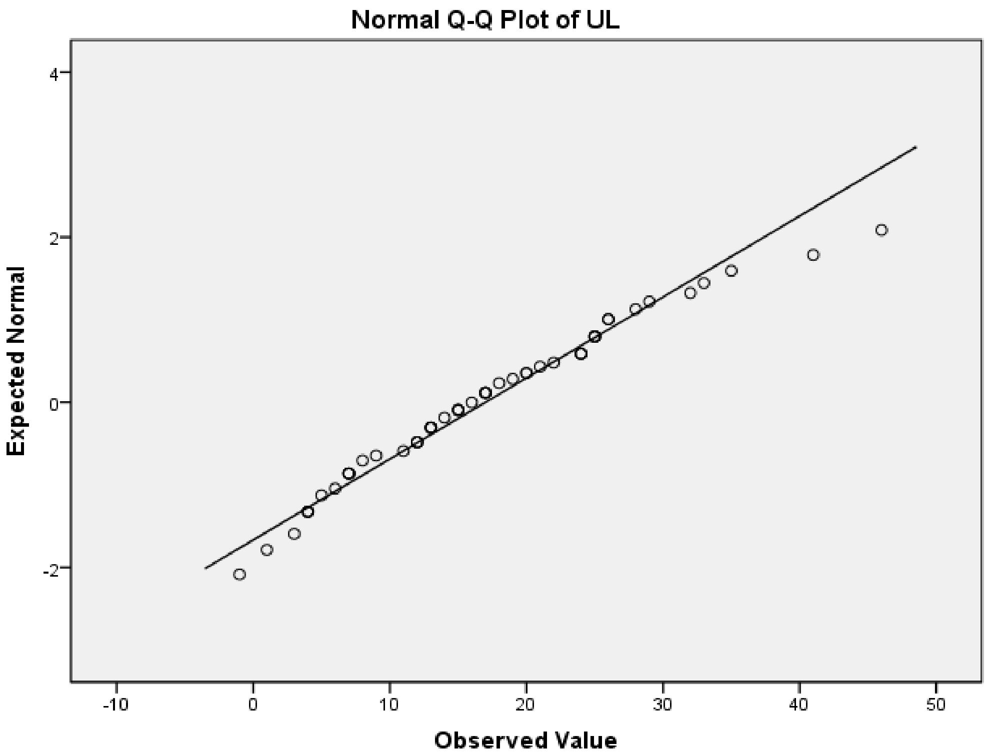
Descriptives

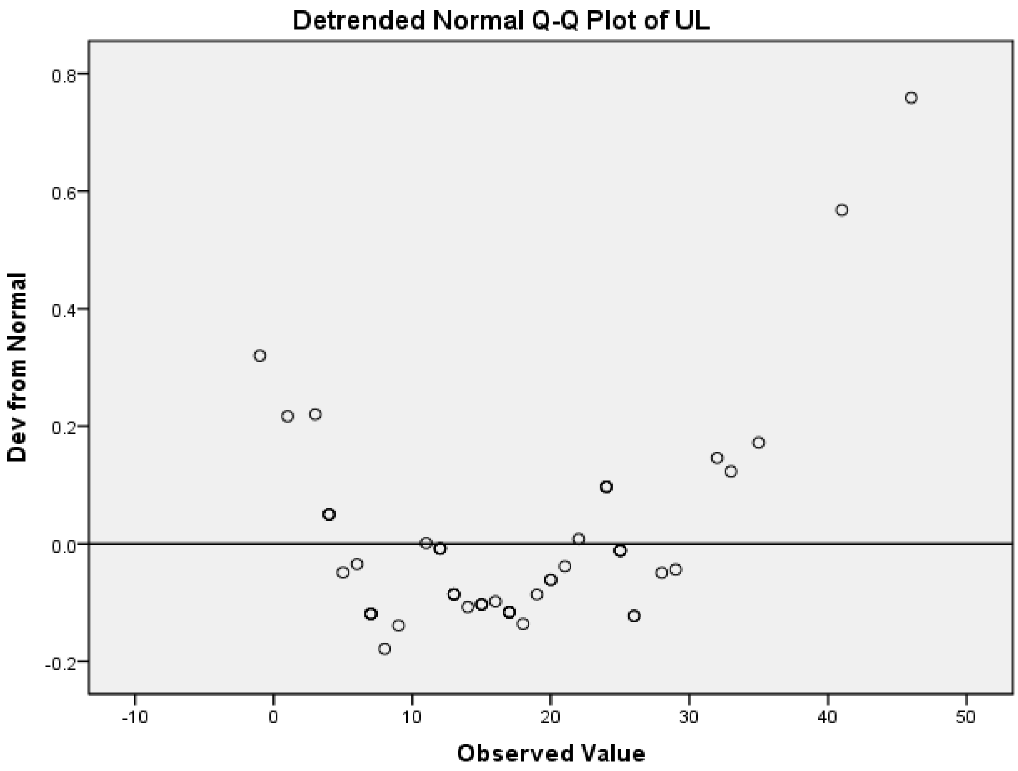
			Statistic	Std. Error
UL	Mean		17.00	1.401
	95% Confidence Interval for Mean	Lower Bound	14.19	
		Upper Bound	19.81	
	5% Trimmed Mean		16.55	
	Median		16.00	
	Variance		103.962	
	Std. Deviation		10.196	
	Minimum		-1	
	Maximum		46	
	Range		47	
	Interquartile Range		16	
	Skewness		.583	.327
	Kurtosis		.250	.644
LL	Mean		27.00	2.072
	95% Confidence Interval for Mean	Lower Bound	22.84	
		Upper Bound	31.16	
	5% Trimmed Mean		26.90	
	Median		27.00	
	Variance		227.500	
	Std. Deviation		15.083	
	Minimum		-12	
	Maximum		59	
	Range		71	
	Interquartile Range		20	
	Skewness		.044	.327
	Kurtosis		.029	.644
WL	Mean		44.00	2.384
	95% Confidence Interval for Mean	Lower Bound	39.22	
		Upper Bound	48.78	
	5% Trimmed Mean		44.01	
	Median		41.00	
	Variance		301.308	
	Std. Deviation		17.358	
	Minimum		7	
	Maximum		85	
	Range		78	
	Interquartile Range		27	

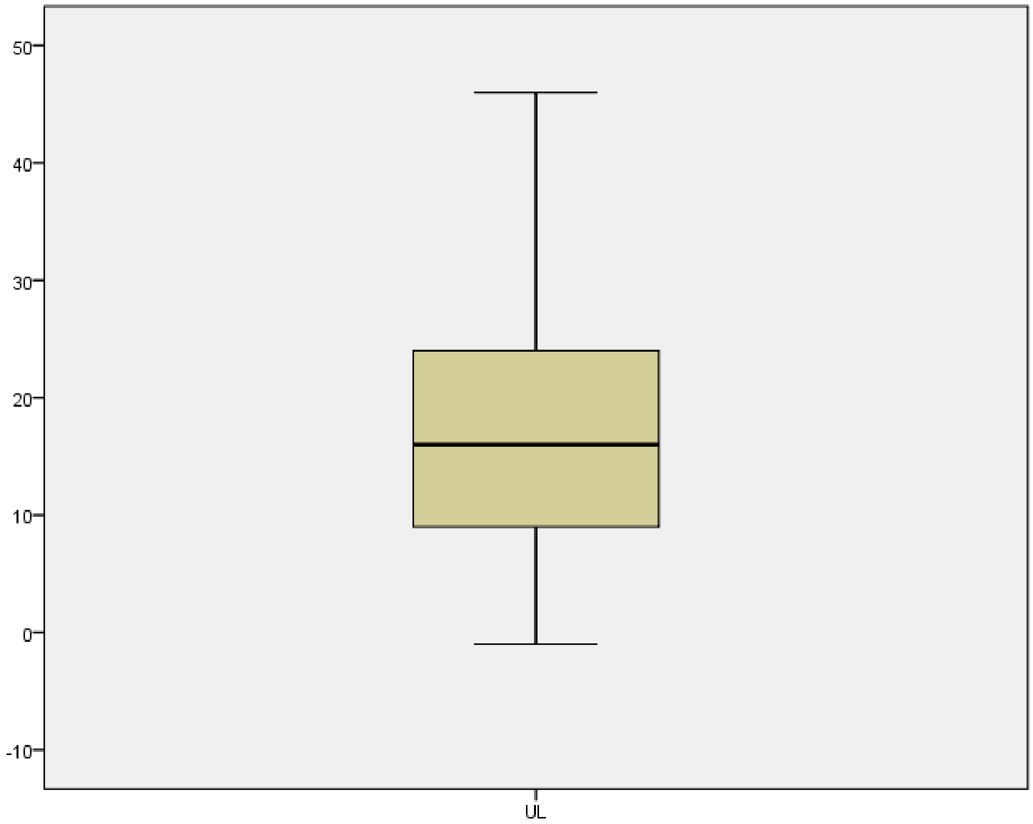
	Skewness	.089	.327
	Kurtosis	-.594	.644

UL

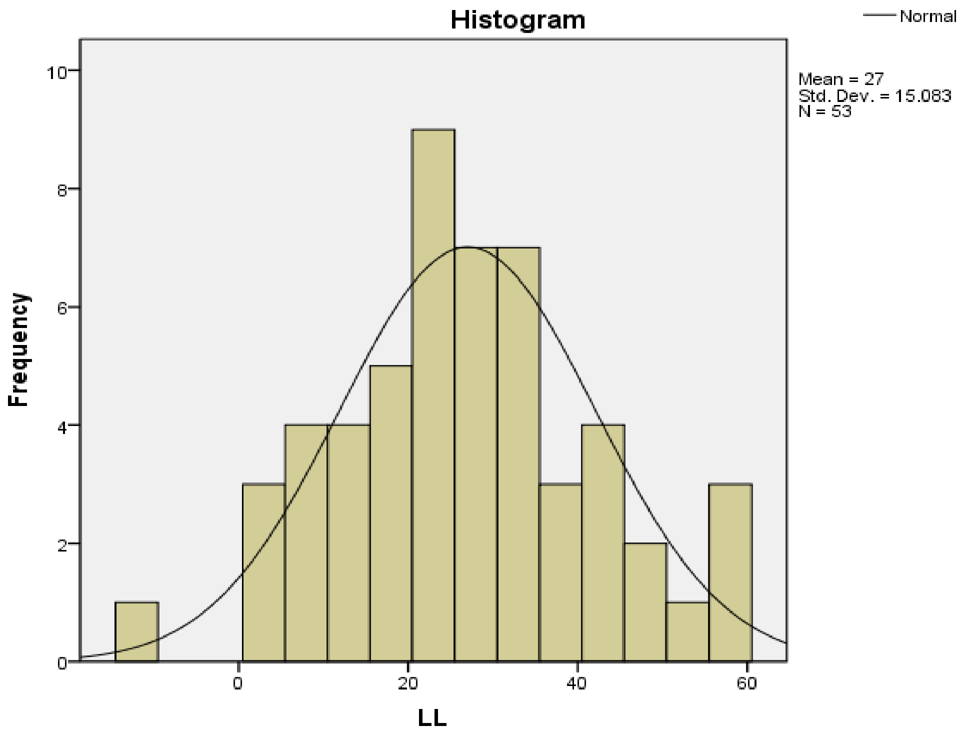


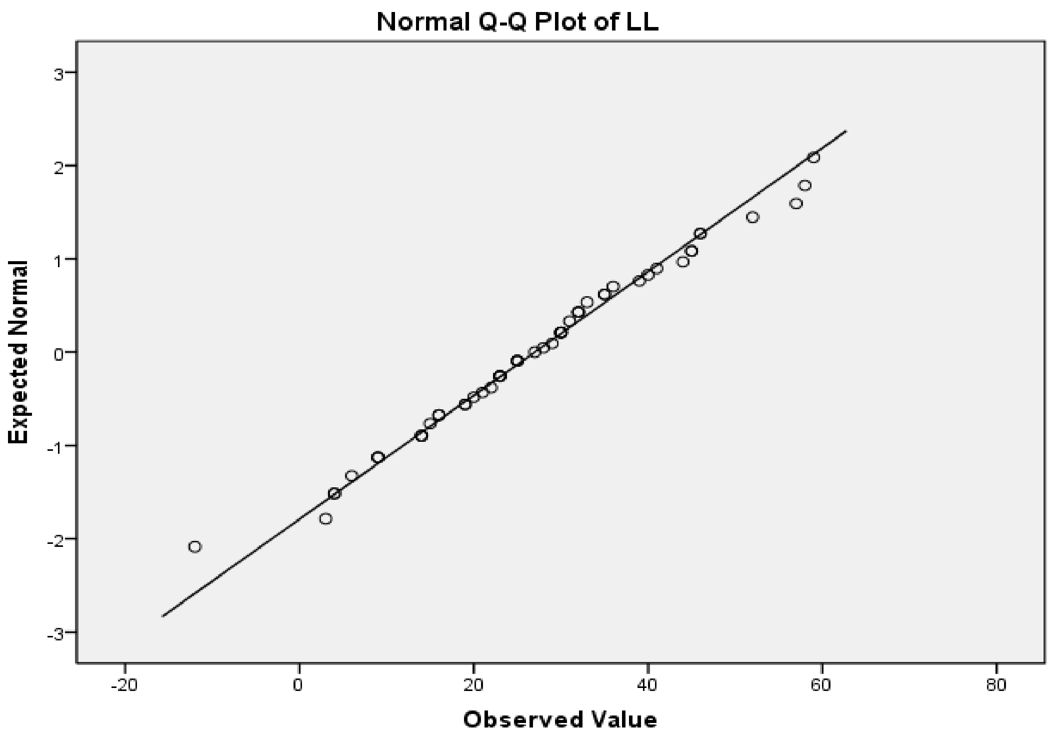

Normal Q-Q Plot of UL
Expected Normal
Observed Value
4
2
0
-2
-10
0
10
20
30
40
50

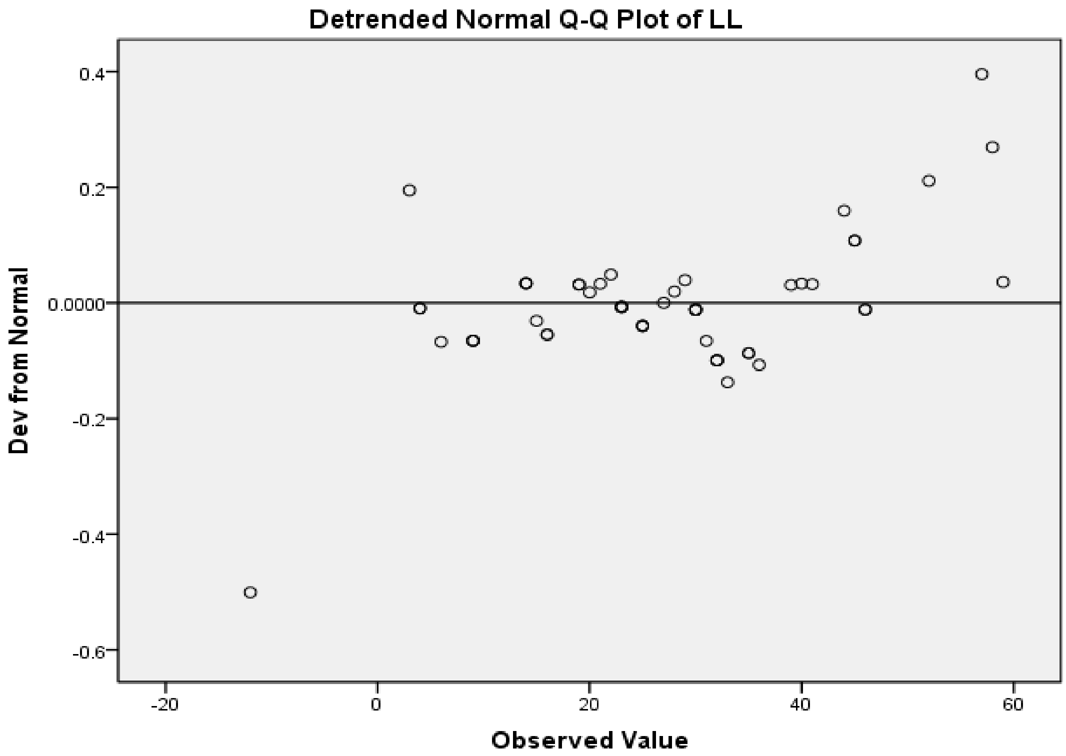

Detrended Normal Q-Q Plot of UL
Dev from Normal
0.8
0.6
0.4
0.2
0.0
-0.2
-10
0
10
20
30
40
50
Observed Value

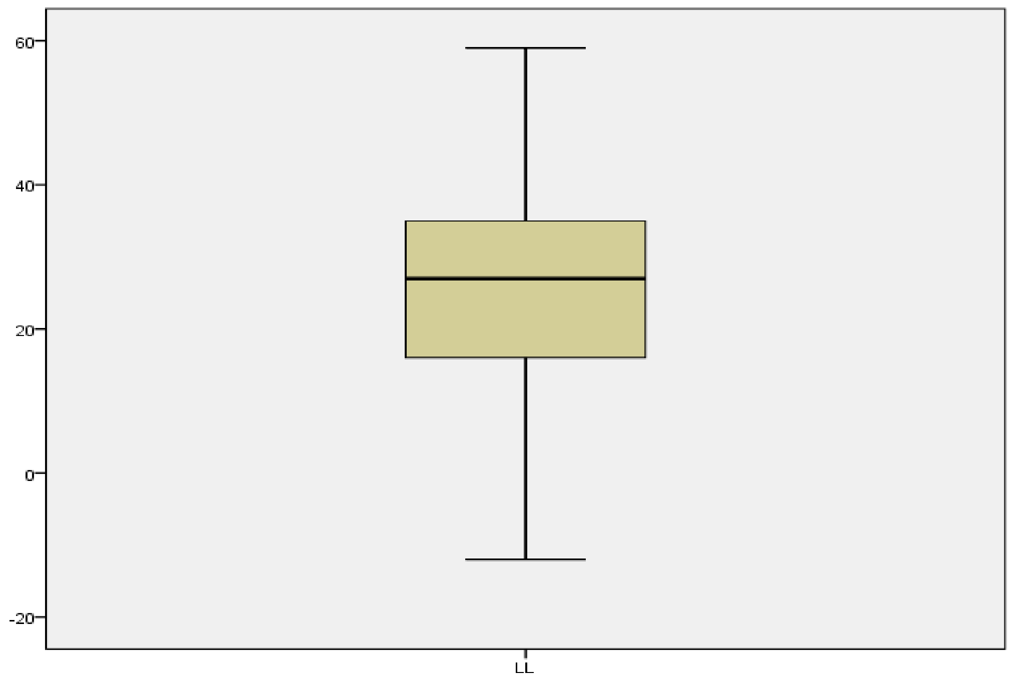

50
40
30
20
10
0
-10
UL

LL

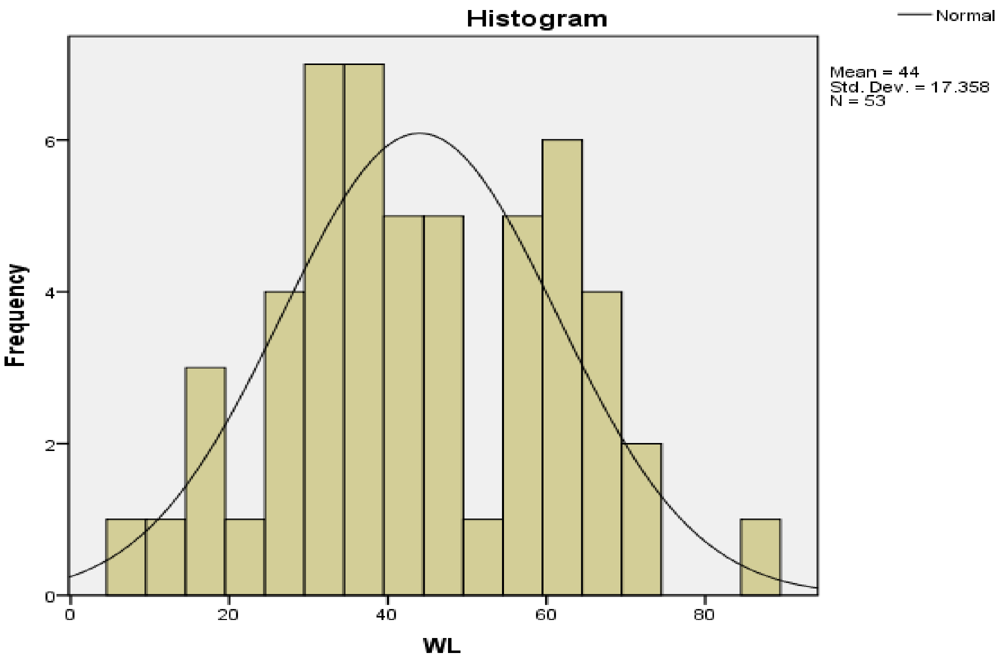

Histogram
Normal
Mean = 27
Std. Dev. = 15.083
N = 53
Frequency
LL
0
2
4
6
8
10
20
40
60


Normal Q-Q Plot of LL
Expected Normal
Observed Value
3
2
1
0
-1
-2
-3
-20
0
20
40
60
80

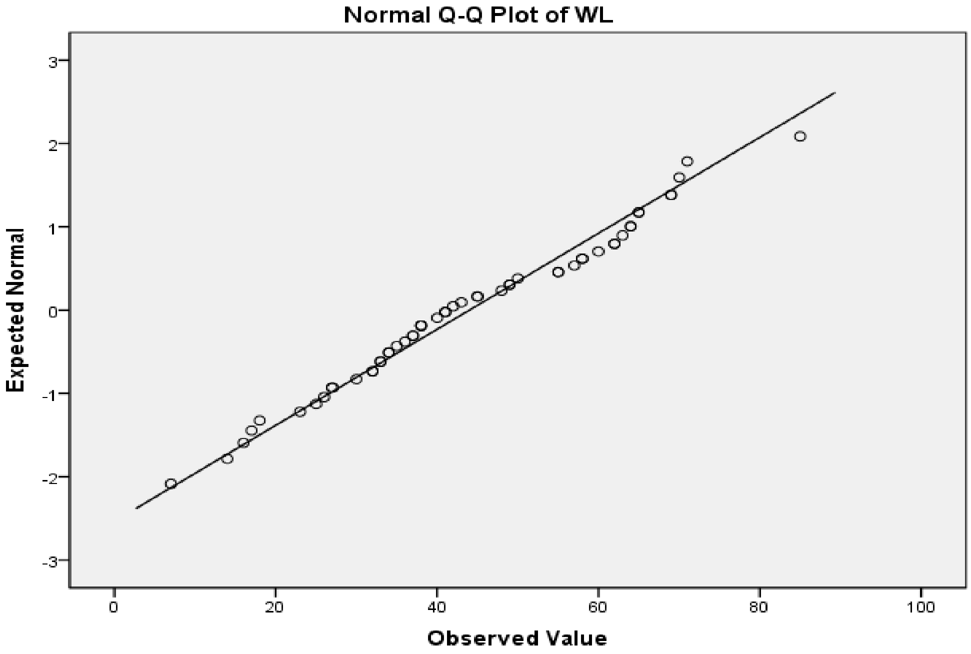

Detrended Normal Q-Q Plot of LL
Dev from Normal
0.4
0.2
0.0000
-0.2
-0.4
-0.6
-20
0
20
40
60
Observed Value

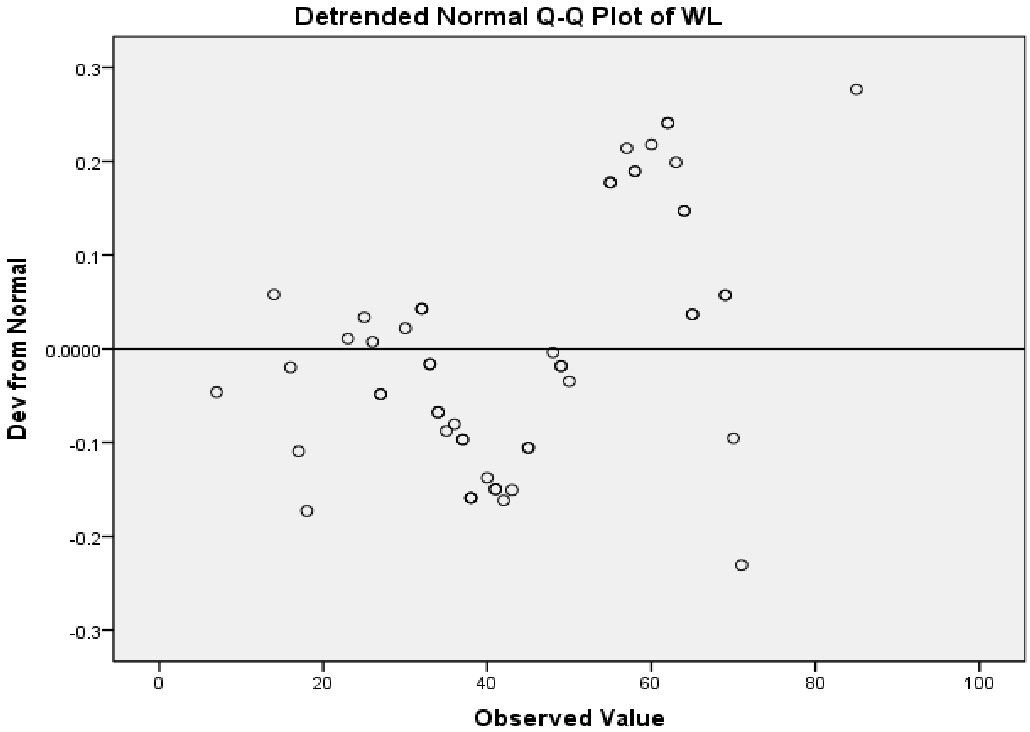

60
40
20
0
-20
LL

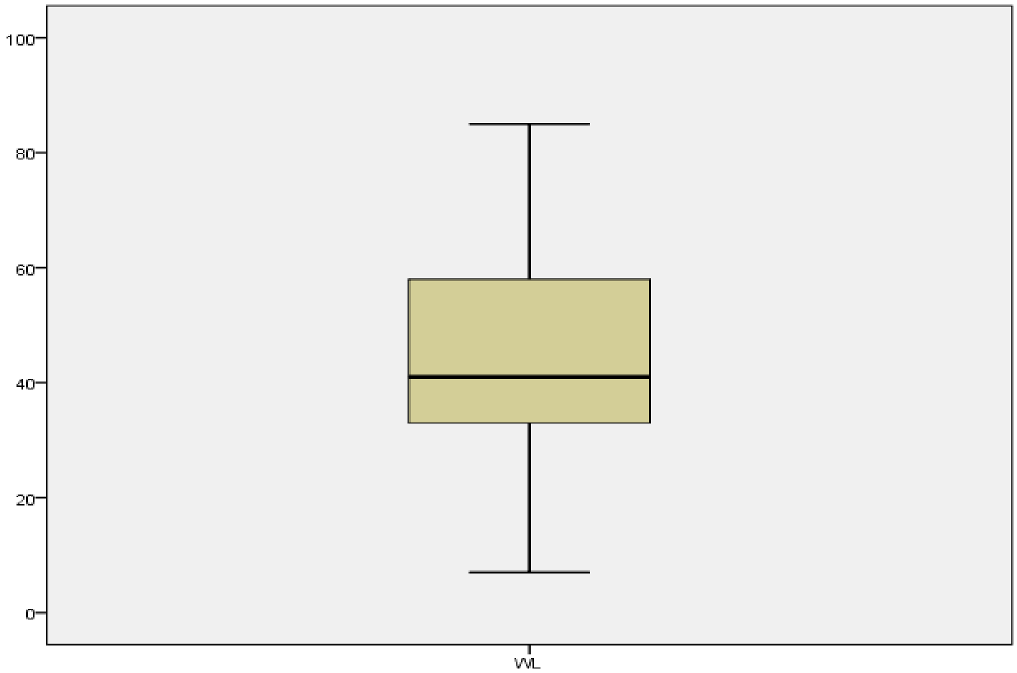
WL



Histogram
Normal
Mean = 44
Std. Dev. = 17.358
N = 53
Frequency
0
2
4
6
0
20
40
60
80
WL


Normal Q-Q Plot of WL
Expected Normal
Observed Value
3
2
1
0
-1
-2
-3
0
20
40
60
80
100


Detrended Normal Q-Q Plot of WL
Dev from Normal
Observed Value
0.3
0.2
0.1
0.0000
-0.1
-0.2
-0.3
0
20
40
60
80
100



v. Siting-standing

Tests of Normality

	Kolmogorov-Smirnov[a]			Shapiro-Wilk		
	Statistic	df	Sig.	Statistic	df	Sig.
UL	.097	53	.200*	.962	53	.089
LL	.072	53	.200*	.990	53	.943
WL	.070	53	.200*	.986	53	.804

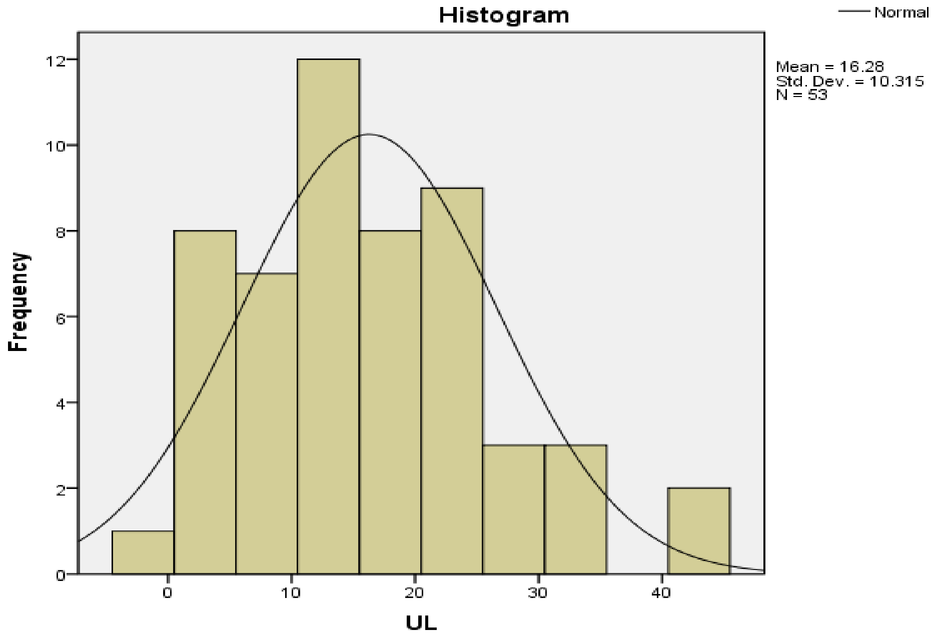
a. Lilliefors Significance Correction

*. This is a lower bound of the true significance.

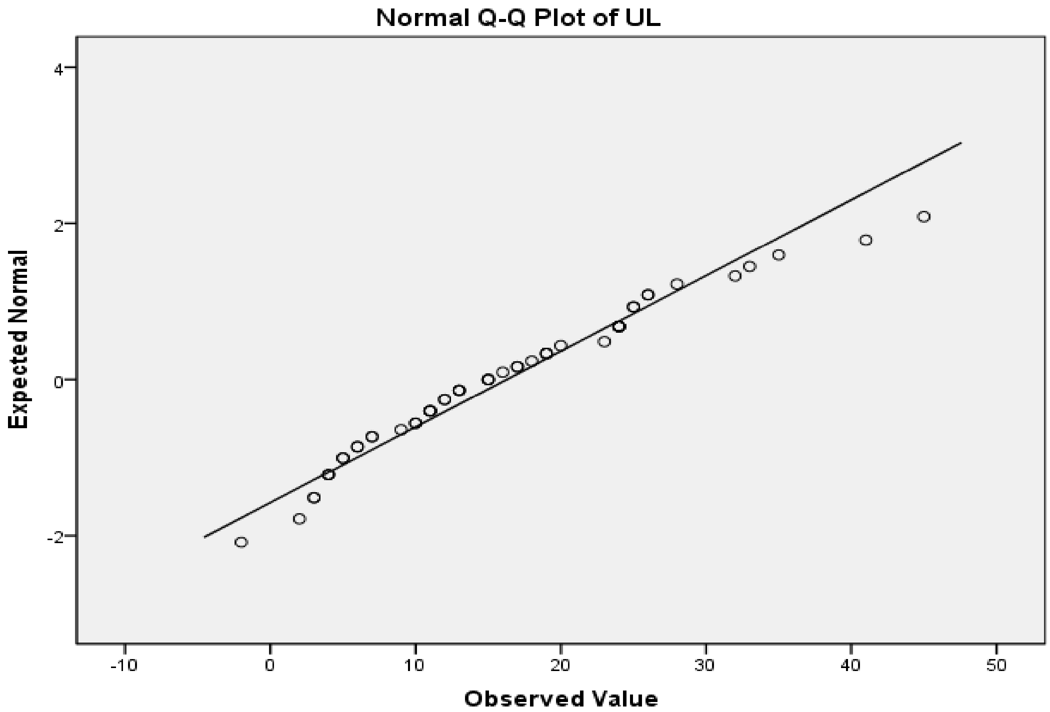
Descriptives

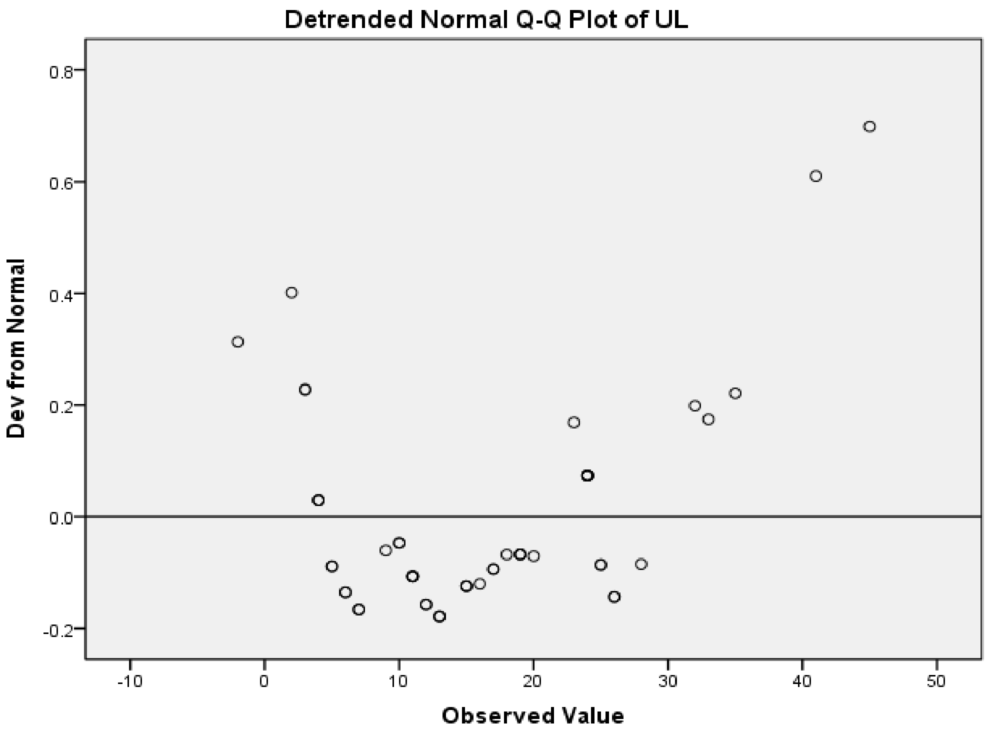
			Statistic	Std. Error
UL	Mean		16.28	1.417
	95% Confidence Interval for Mean	Lower Bound	13.44	
		Upper Bound	19.13	
	5% Trimmed Mean		15.77	
	Median		15.00	
	Variance		106.399	
	Std. Deviation		10.315	
	Minimum		-2	
	Maximum		45	
	Range		47	
	Interquartile Range		16	
	Skewness		.622	.327
	Kurtosis		.157	.644
LL	Mean		26.64	2.066
	95% Confidence Interval for Mean	Lower Bound	22.50	
		Upper Bound	30.79	
	5% Trimmed Mean		26.70	
	Median		25.00	
	Variance		226.157	
	Std. Deviation		15.039	
	Minimum		-10	
	Maximum		61	
	Range		71	
	Interquartile Range		21	
	Skewness		-.008	.327
	Kurtosis		-.384	.644
WL	Mean		43.98	2.555
	95% Confidence Interval for Mean	Lower Bound	38.85	
		Upper Bound	49.11	
	5% Trimmed Mean		43.82	
	Median		42.00	
	Variance		346.019	
	Std. Deviation		18.602	
	Minimum		4	
	Maximum		84	
	Range		80	
	Interquartile Range		28	
	Skewness		.130	.327
	Kurtosis		-.405	.644

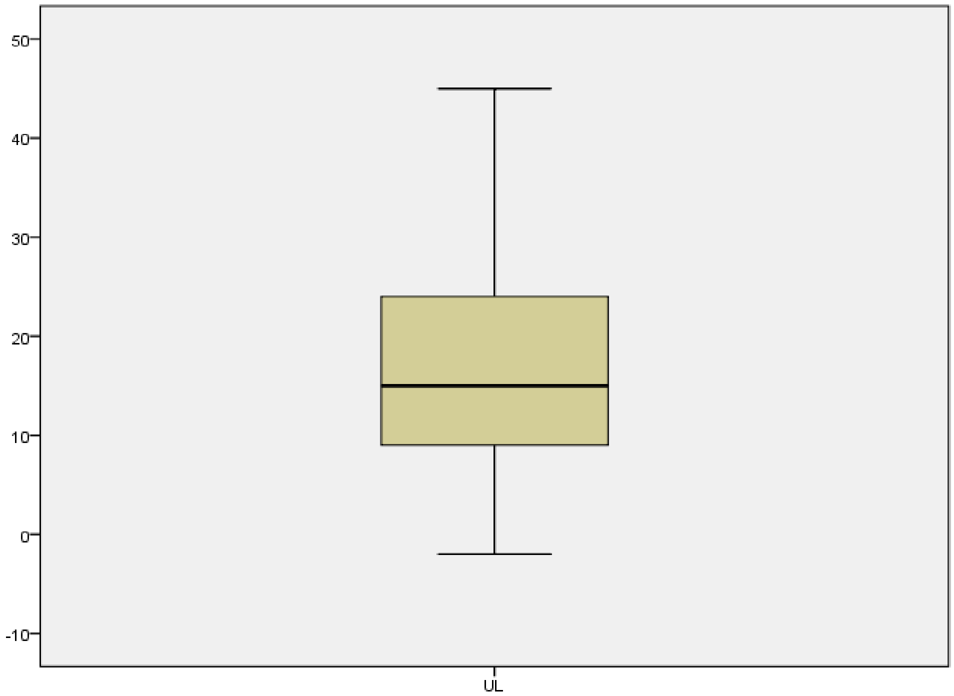
UL



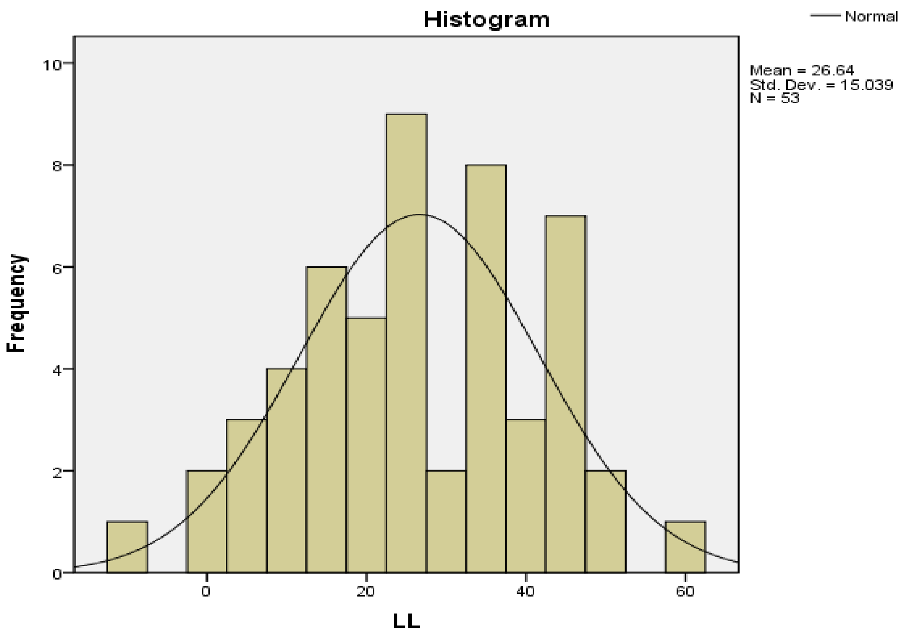
Histogram
Normal
Mean = 16.28
Std. Dev. = 10.315
N = 53
Frequency
12
10
8
6
4
2
0
0
10
20
30
40
UL

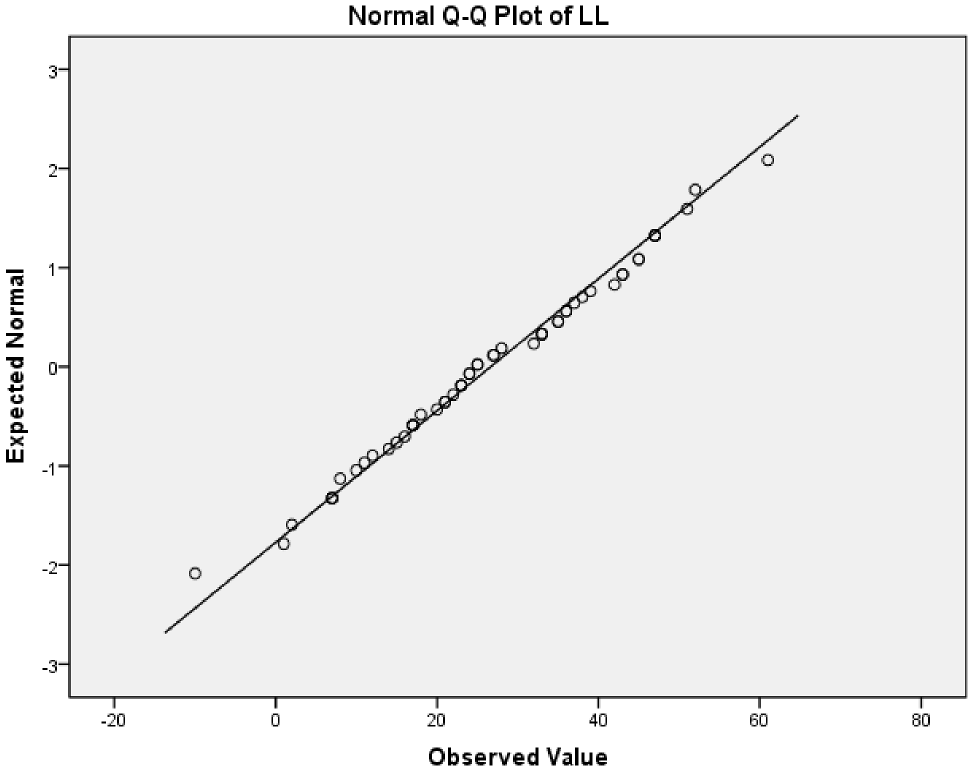

Normal Q-Q Plot of UL
Expected Normal
Observed Value
4
2
0
-2
-10
0
10
20
30
40
50


Detrended Normal Q-Q Plot of UL
Dev from Normal
Observed Value
0.8
0.6
0.4
0.2
0.0
-0.2
-10
0
10
20
30
40
50

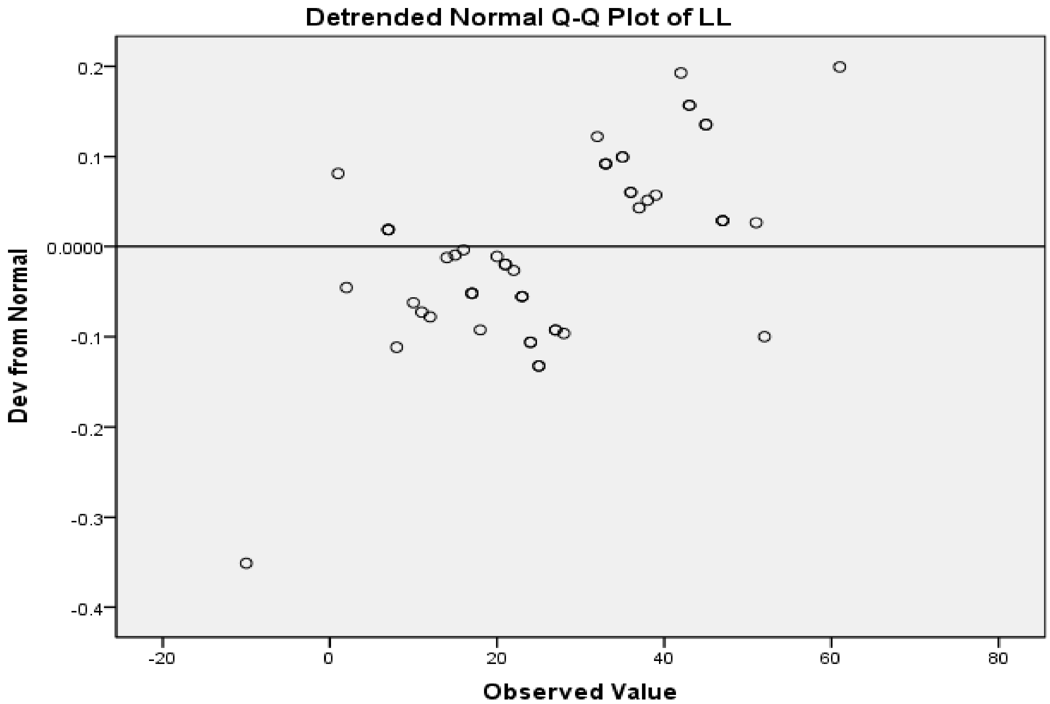

50
40
30
20
10
0
-10
UL

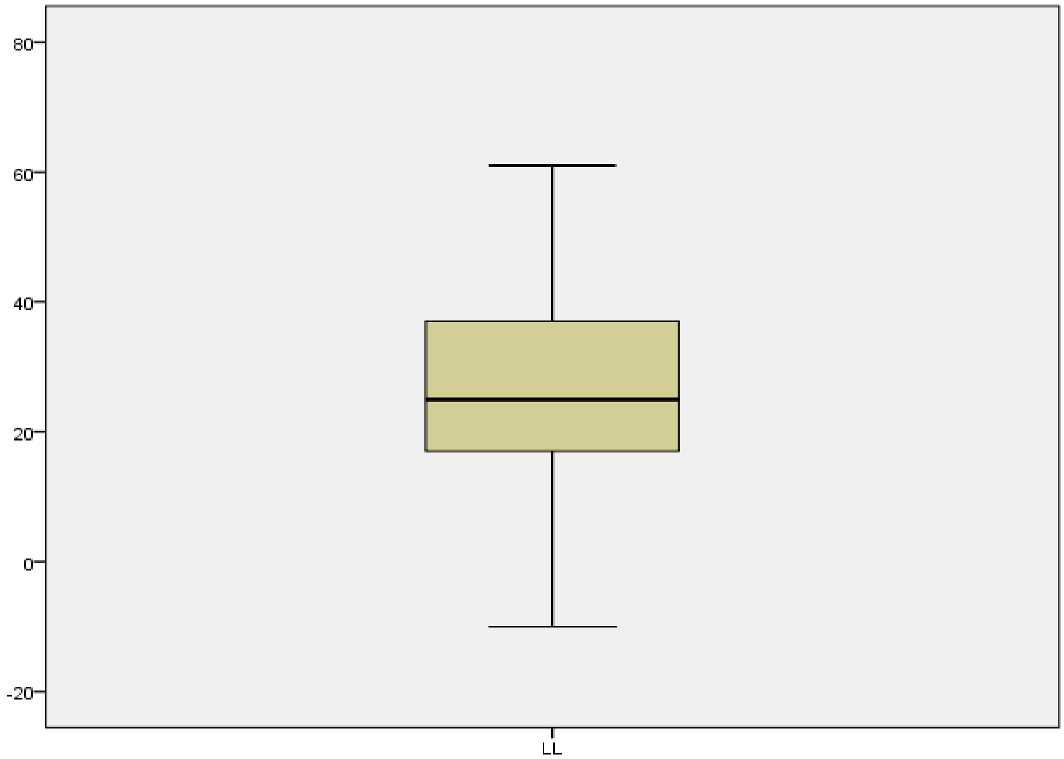
LL


Histogram
Normal
Mean = 26.64
Std. Dev. = 15.039
N = 53
Frequency
LL

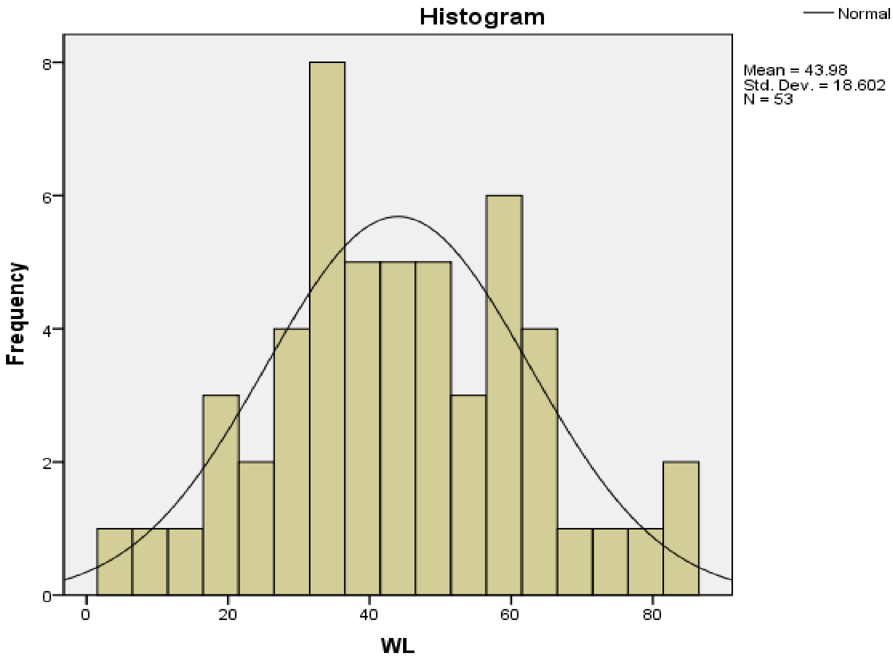


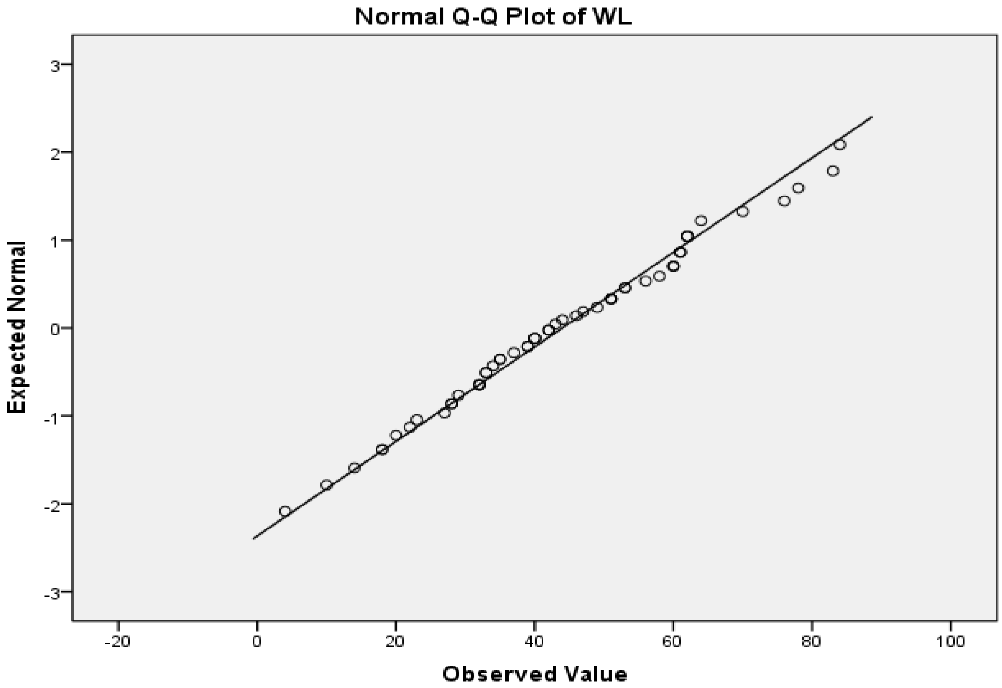
Normal Q-Q Plot of LL
Expected Normal
Observed Value
3
2
1
0
-1
-2
-3
-20
0
20
40
60
80


Detrended Normal Q-Q Plot of LL
Dev from Normal
Observed Value
0.2
0.1
0.0000
-0.1
-0.2
-0.3
-0.4
-20
0
20
40
60
80

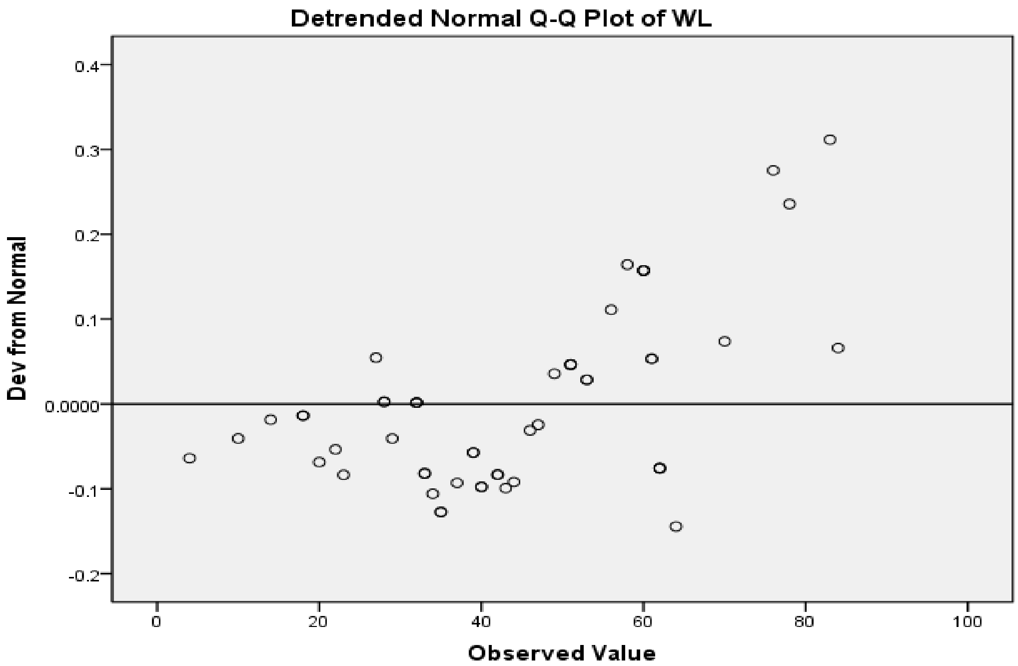

80
60
40
20
0
-20
LL

WL

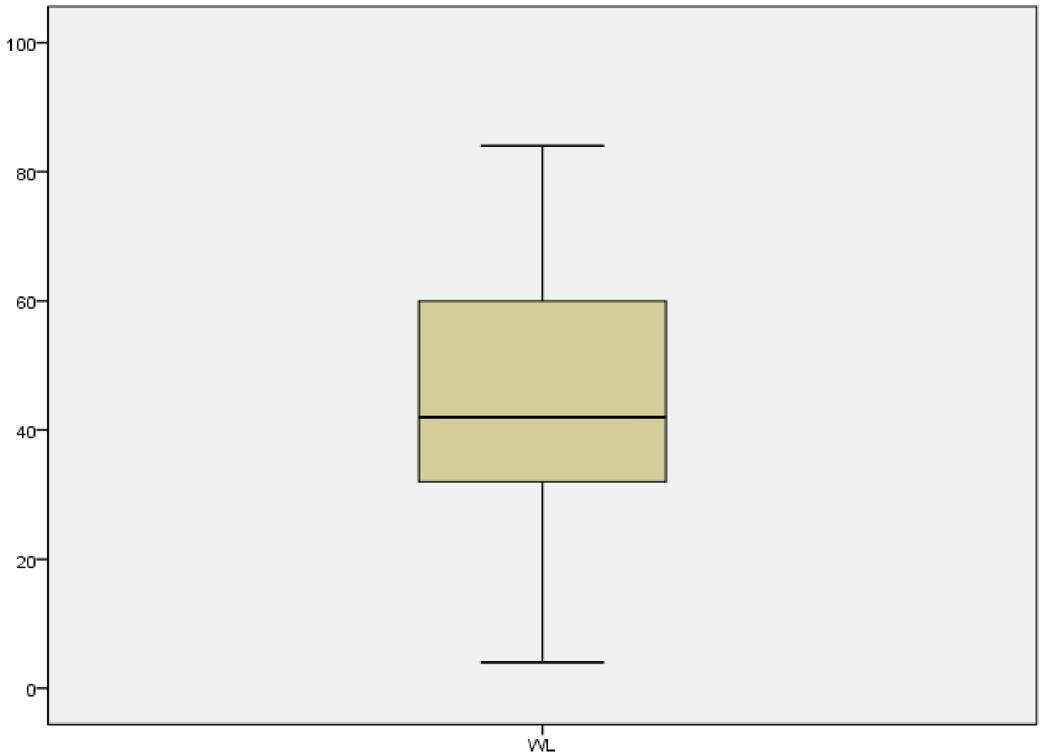

Histogram
Normal
Mean = 43.98
Std. Dev. = 18.602
N = 53
Frequency
0
2
4
6
8
0
20
40
60
80
WL



Normal Q-Q Plot of WL
Expected Normal
Observed Value
3
2
1
0
-1
-2
-3
-20
0
20
40
60
80
100



Detrended Normal Q-Q Plot of WL
Dev from Normal
Observed Value
0.4
0.3
0.2
0.1
0.0000
-0.1
-0.2
0
20
40
60
80
100



100
80
60
40
20
0
WL