DOCTORAL THESIS

Vibration transmission through the human spine during physical activity

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Award date:
2013

Awarding institution:
University of Roehampton

Vibration transmission through the human spine during physical activity

by

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A thesis submitted in partial fulfilment of the requirements of the degree of PhD

Department of Life Sciences

University of Roehampton

2013

Abstract

Osteoporosis causes bone to become fragile. Pharmacological treatments of osteoporosis are burdened with adverse effects and increase bone mineral density (BMD) only between 1% and 15% depending on the drug and time used. Thus non pharmacological treatments are needed to complement pharmacological ones. Physical activity is a non pharmacological treatment of osteoporosis and is essential for maintaining bone health at any age. However, physical activities have been identified to produce a modest improvement of spinal strength or just preserve it. In addition, it is not known how much exercise is optimal and safe for people with spinal osteoporosis. Most research employs conflicting definitions of physical activity and measure the effect of exercise on BMD alone instead of combining it with measurements of three dimensional bone strength. There is the need to offer a technique to measure the effect of physical activity on the overall strength of the spine, not only on its bone mineral content.

Vibration transmissibility is a measurement of the mechanical response of a system to vibration expressed as stiffness or damping, thus offering a variable that represents structural strength. It can be employed to measure the mechanical response of the human spine during physical activity by attaching inertial sensors over the spine. However, it has not been employed to characterize the way vibration is transmitted through the osteoporotic spine during physical activity. Understanding the effects of osteoporosis and ageing on vibration transmission is important since such effects are related to the stiffness of the spine and thus very likely to the incidence of vertebral fractures. It is also often recommended that fast walking is beneficial to the bone, yet it is not known if fast walking affects the mechanical response of the spine of people with osteoporosis.

The aims of this study were (1) to evaluate the feasibility of employing inertial sensors and a skin correction method to measure vibration transmission through the spine during physical activity (2) to characterize the transmission of vibration in the lumbar and thoracic spines of people with and without osteoporosis during physical activities, (3) to characterize the effect of osteoporosis on

vibration transmissibility at levels of the thoracic spine which are known to fracture and (4) to investigate the effects of fast walking on vibration transmissibility.

100 young and healthy and older volunteers with and without osteoporosis were recruited. Participants were asked to perform straight walking, stair negotiation and turning while having inertial sensors attached to the skin over the spinous process of the first sacral (S1), twelfth (T12), eighth (T8) and first thoracic vertebrae (T1). Vibration transmissibility was calculated as the square root of the acceleration of the output (T12 for the lumbar and T1 for the thoracic spine) over the input (S1 for the lumbar and T12 for the thoracic spine) in the frequency spectrum. Vibration transmissibility was corrected for the movement of the skin-sensor interface and for the inclination of the sensor over the spine of every subject. All physical activities were performed at self selected normal and fast walking speeds. Lumbar and thoracic curvatures were determined with an electromagnetic device and BMD was measured through quantitative ultrasound.

Skin measurement of transmission of vertical vibration is feasible with the inertial sensors and correction method presented. Vibration transmissibility through the human spine is significantly different between dissimilar physical activities and frequency dependent. Ageing significantly alters the vibration transmissibility of the spine. Osteoporosis has a minimal effect on vibration transmissibility of the spine. The effect of ageing and osteoporosis are frequency dependent. Older lumbar spines may receive greater stimulation than young and healthy ones, whereas older thoracic spines may receive lower stimulation during fast walking. There are significant differences in vibration transmissibility between lumbar and thoracic spines. A percentage of vibration transmission of the lumbar and thoracic spines is determined by their curvatures.

This thesis has provided a technique that future research can employ to correlate vibration transmissibility with mechanotransduction signals in bone as well as volumetric bone health measurements and the risk of vertebral fractures. Until then it will be possible to prescribe physical activity taking into account individual capabilities, bone strength and differences in mechanical response between lumbar and thoracic sections.

Contents

Acknowledgments

Gracias a Gela, Vladi y Gus por la paz que me han dado cuando la he necesitado, gracias por recordarme que hay una familia apoyándome y amándome incondicionalmente. Los amo.

Bahat shukria Basit. Thank you for not letting me work alone (sometimes), thank you for your eternal patience and the long suffering hours you spent waiting and holding my stressful soul through this journey. I love you.

Thanks to my friends for their emotional support, for their multiple perspectives and for continue talking to me after this PhD isolation.

Special thanks to Raymond Lee and Siobhan Strike for their support.

Thanks to Alison and Tom for their technical support in the Biomechanics laboratory.

I am forever grateful to University of Roehampton for their funding during this research project.

Gracias a la vida que me ha dado tanto.

"Everything is connected"

Declaration

I confirm that the contents of this report are entirely my own work and that nothing has been included from other sources without acknowledgement of reference.

Publications

MORGADO RAMÍREZ, D. Z., STRIKE, S. & LEE, R. Y. W. 2013. Measurement of transmission of vibration through the human spine using skin-mounted inertial sensors. Medical Engineering & Physics, 35, 690-695

MORGADO RAMÍREZ, D. Z., LEE, R. & STRIKE, S. 2013. Osteoporosis and aging alter the vibration transmission of the spine during physical activity: the lumbar spine (Part I). Osteoporosis International, under review

MORGADO RAMÍREZ, D. Z., LEE, R. & STRIKE, S. 2013. Osteoporosis and aging alter the vibration transmission of the spine during physical activity: the thoracic spine and implication of osteoporosis (Part II). Osteoporosis International, under review

Conference Proceedings

MORGADO RAMIREZ, D. Z., LEE, R. & STRIKE, S. Surface measurement of vibration transmissibility of the human spine during daily life activities. In: ISB Congress XXIII, 2011 Brussels, Belgium. International Society of Biomechanics.

Manuscript in preparation

MORGADO RAMÍREZ, D. Z., LEE, R. & STRIKE, S. 2013 Fast walking alters vibration transmission of the spine in healthy and osteoporotic subjects. Journal of Biomechanics, to be submitted in July 2013

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List of Abbreviations

ζ	Damping ratio
fn	Natural frequency
PSD	Power spectral density
FFT	Fast Fourier transform
ASD	Acceleration spectral density
w	Straight walking
m	Combined walking and turning
a	Ascending stairs
d	Descending stairs
NWS	Self selected normal walking speed
FWS	Self selected fast walking speed
SD	Standard deviation
RMSa	Root Mean Square Acceleration
YH	Young and Healthy
OH	Older Healthy
OO	Older Osteopenic and Osteoporotic
WS	Walking speed
FEA	Finite Element Analysis
με	Micro strain
Hz	Hertz
T1	First thoracic vertebra
T8	Eighth thoracic vertebra
T12	Twelfth thoracic vertebra
S1	First sacral vertebra
≈	Almost equal to
IU	International unit

CHAPTER 1 Introduction

1.1. Statement of the problem

1.1.1. Osteoporosis

Osteoporosis is a major public health problem which affects the lives of a considerable number of older adults. It often leads to painful fractures, causing increased mortality and reduced quality of life. The incidence of vertebral fractures worldwide, which were related to osteoporosis, was found to be 1.4 million per annum. About 39% of these fractures occurred in men and 60% in women (Johnell and Kanis, 2006). It is estimated that 34% of vertebral fractures are under diagnosed worldwide (IOF, 2012). Vertebral fractures have a significant impact on daily life activities since they cause back pain, loss of height, deformity, immobility and reduced pulmonary function (IOF, 2012). The cost of osteoporotic fractures in Europe is over €36 billion annually and it is expected to increase to €77 billion by 2050 (IOF, 2012).

Current pharmacological treatments of osteoporosis increase spinal BMD between 1% and 15% and reduce vertebral fracture risk between 30% and 83% (Burr et al., 2002, NOF, 2010, Amrein et al., 2012, Woo and Adachi, 2006). Treatment of osteoporosis with drugs alone is not sufficiently efficient, non pharmacological interventions as complementary treatments are needed. One non pharmacological treatment that offers great opportunities is physical activity.

1.1.2. Physical activity and walking speed

Common physical activities in our everyday life involve walking in a straight line, stair negotiation and turning. However these represent a unique challenge to older adults as functional impairment, pain and limited range of motion are associated with osteoporosis of the spine. Although physical activity is generally believed to be beneficial and may help increase or maintain BMD, it may also increase the risk of vertebral fractures in osteoporotic patients, and outweigh its benefits. The consistent effectiveness of exercise to increase BMD or at least impede its loss has not been demonstrated in large randomised controlled trials (Bergmann et al., 2011, Kasturi and Adler,

2011, Hamilton et al., 2010a, Cheung and Giangregorio, 2012, Gremeaux et al., 2012, Mayer et al., 2011, Kohrt et al., 2004). Current physical activity prescription for people with osteoporosis is based on casual observations or indications rather than rigorous scientific analysis. Similarly, it has been suggested that increasing the speed of walking may improve bone health. However, there is no agreement on the effect of fast walking since there are studies recommending it (Winter-Stone, 2005, NOF, 2010, Van Norman, 2010, NOF, 2012) while others have found no effects on spinal BMD (Martyn-St James and Carroll, 2008, Schmitt et al., 2009). Thus it is not clear if the effect of physical activity on the mechanical response of the spine is dependant of walking speed.

The conceptual framework of this thesis is based on the fact that the effectiveness and the effect of physical activity and walking speed have not been measured with a standard method across different research studies. There is no standard way to classify physical activity thus current methods lead to overlap of research results across studies attempting to understand the effect of physical activity on skeletal strength among people with osteoporosis. This suggests that there is the need to find a practical technique to measure the effect of exercise on bone noninvasively while providing a measurement related not only to BMD but to bone size and structure as the mayor contributors of skeletal strength.

1.1.3. Assessment of spinal osteoporosis

Medical imaging techniques, which are still in research and development, are capable of assessing the strength of the human skeleton (WHO, 2003, Bauer and Link, 2009, Griffith and Genant, 2012, Griffith et al., 2006, Techawiboonwong et al., 2008, Christiansen et al., 2011, Ahmad et al., 2010, Wu et al., 2010, Schmidt et al., 2007, Schmidt et al., 2006, Schmidt et al., 2010, Cheung et al., 2003, Niemeyer et al., 2012). However, they measure strength given a static posture and are only available at hospital level. Given that medical imaging based techniques to measure bone strength are not portable there is the need to develop a technique which should be movable and be able to provide a measurement of bone strength while performing physical activity.

Vibration transmissibility is a technique capable of providing a measurement which describes the stiffness and compliance of any structure (Mansfield, 2005a). It consists in the attachment of inertial sensors to the spine over specific spinous processes and therefore enables its operation outside a hospital (Helliwell et al., 1989, Smeathers, 1989a, Smeathers, 1989b, Kitazaki and Griffin, 1995). Vibration transmissibility has the advantage of considering BMD, tissue properties and structure. Since vibration is produced and transmitted though the body during gait (Cappozzo, 1982, Voloshin et al., 1981), vibration transmissibility becomes an ideal technique to characterize the mechanical response of the spine to physical activity at different walking speeds and for people with and without osteoporosis.

The spine may be able to attenuate and amplify vibration produced during daily physical activities and transmitted to the spine. The way this vibration is transmitted through the spine may be altered by ageing and osteoporosis. Physical activity may not have the same effects on the vibration transmissibility across individuals of different ages and different bone health. Thus present physical activity prescriptions may be unsafe or have no significant effect on the treatment of spinal osteoporosis. At present there is no information in the way vibration is transmitted through the spine during physical activity. It is hoped that this study will increase our understanding of the nature of vibration transmitted through the spine. This will allow us to understand how physical activity may affect the mechanical response of the spine. It is expected that this study will clarify the effect of osteoporosis and ageing on the mechanical response of the spine during physical activity through the measurement of vibration transmissibility.

1.2. Purpose of the study

The main aim of this study was to characterize and analyse the mechanical signals that are transmitted through the human lumbar and thoracic spines (healthy and osteoporotic) while performing daily life physical activities at two different walking speeds.

1.2.1. Objectives

1. To evaluate the feasibility of employing inertial sensors for spinal vibration transmission measurement during physical activity.
2. To study the effect of different physical activities, ageing and osteoporosis on vibration transmission of the spine.
3. To study the characteristics of vibration transmission by comparing the lumbar and thoracic spines.
4. To study the effect of walking speed on vibration transmission of the spine.
5. To study the relationship between spinal curvature and vibration transmission through the spine.

1.3. Scope and boundaries of thesis

It is within the scope of this study to evaluate the feasibility of employing the vibration transmissibility measurement, based on inertial sensors, and to characterize the effect of various physical activities on the healthy and osteoporotic human spine at different walking speeds. The development of a device ready to be used in the clinic to aid clinicians to prescribe physical activity individually is not within the scope of this thesis. However, this study offers the first step towards the development of such device by providing evidence of the usefulness of vibration transmissibility for characterizing the effect of physical activity on the mechanical response of the spine of individuals with and without osteoporosis. Similarly, it is within the scope of this thesis to characterize the effect of two different walking speeds on vibration transmissibility of the spine in a sample of the population with and without osteoporosis.

The types of physical activities performed in our daily life vary across individuals. The physical activities tested in this study are a sample that is considered to be representative of basic daily activities. It is not within the scope of this thesis to measure the effect of physical activities on spinal strength over time, for example monitoring during a year. Instead, the effects of specific physical activities are measured in a single session, during a single day for each individual representing either a young and healthy group or an older healthy or osteoporotic group.

The study of the anatomy and physiology of the human spine is not within the scope of this thesis. Similarly, the reader is expected to have basic knowledge of the biomechanics of the healthy human spine.

A representative group of the population with and without osteoporosis was recruited. A specific inclusion criteria was employed in this study in order to include only those subjects without any life event that could have modified the biomechanics of the spine in an unnatural way (anything different to ageing and osteoporosis). Any cause of injury or disease meant subjects were excluded from this study.

This study employed inertial sensors which are portable and enable the volunteers to perform physical activity free from any physical restriction. However, due to the intrinsic limitations of the physical principles in which the inertial sensors function, the characterization of the vibration transmitted by the spine was only done in the vertical direction.

Specific boundaries are presented within each chapter.

1.4. Outline of thesis

Chapter 2 provides a review of the current treatment of osteoporosis focusing on the effectiveness of prescribed physical activity. The employment of a method (based in inertial sensing technology) to study the mechanical response of the spine during physical activity is suggested due to its advantages over other osteoporosis monitoring techniques. Chapter 3 examines the feasibility of measuring the transmission of vibration through the human spine using skin mounted inertial sensors. The effect of corrections for skin movement and sensor inclination are objectively established. Chapter 4 studies the feasibility of using vibration transmission to identify the effect of ageing and osteoporosis on the lumbar spine during different physical activities. Chapter 5 explores if vibration transmission, to locations of the thoracic spine where vertebral fractures are common, is significantly affected by ageing and osteoporosis. Differences between lumbar and thoracic spines are also presented. Having established the biomechanical response of the spine to different physical activities and the effect of aging and osteoporosis, Chapter 6 studies the effect of increasing

walking speed on that biomechanical response. Chapter 7 presents a general discussion, clinical interpretation, application and limitations of the vibration transmission measured.

1.5. Definition of terms

Vibration: mechanical movement that oscillates about a fixed point. Mechanical wave that transfers energy through a structure (or person) (Mansfield, 2005a).

Spectral analysis: process by which signals (time series) are converted from the time domain to the frequency domain (Mansfield, 2006).

Fourier Series: any periodic waveform can be represented as the sum of an infinite number of sinusoidal and cosinusoidal terms, which with a constant term represent the Fourier Series (Ifeachor and Jervis, 2002).

Fourier Transform (FT): when waveforms are not periodic the Fourier series modified (Ifeachor and Jervis, 2002).

Fast Fourier Transform (FFT): stands for the algorithm for the fast computation of the Fourier Transform (Ifeachor and Jervis, 2002).

Power Spectral Density: is the average power (energy) in a signal (time series) over a frequency interval. The term density is used because the power in each selected frequency interval has been divided by the width of the interval (Griffin, 1990). It splits the original signal into shorter segments and calculates the FFT of each section. The units are $(m/s^2)^2/Hz$ (Mansfield, 2005a)

Frequency Response Function: the frequency response of a discrete time system is the Fourier transform of its impulse response (Ifeachor and Jervis, 2002). The characteristics of the response of a system, to vibration at any frequency, are calculated using frequency response (transfer) functions. If at any frequency the magnitude of the input and output (of the measured system) are identical, the transfer function is unity.

Coherence: extent of correlation between an input and output signal. Powerful tool for providing an indication of the reliability of a measurement (Mansfield, 2005a).

Compliance: reciprocal of a system's stiffness.

Ankylosing spondylitis: chronic painful inflammation of spinal joints which causes spinal stiffness.(Helliwell et al., 1989)

Sarcopenia: loss of muscle mass and strength with aging (Morley et al., 2001).

Proteoglycans: proteins associated with glycosaminoglycans (GAGs), they trap water enabling the jellylike properties of the ground substance in the matrix of connective tissues (Tortora and Grabowski, 2003).

Osteomalacia: disease in which bone formed during remodelling fails to calcify. Causes varying degrees of pain and fractures after minor trauma (Tortora and Grabowski, 2003).

Catecholamine: any of various amines that function as neurotransmitters and hormones (dopamine, adrenaline and noradrenalin) (Schulz et al., 2004).

Progenitor cells: myeloid stem cells differentiated during hemopoiesis. These are no longer capable of reproducing themselves but committed to give rise to specific elements of blood (Tortora and Grabowski, 2003).

Diffusion: movement of molecules from an area of higher concentration to another area of lower concentration (Tortora and Grabowski, 2003).

Perfusion: injection of fluid into a blood vessel in order to reach a specific tissue to supply nutrients and oxygen (Tortora and Grabowski, 2003).

Feldenkrais: educational method focusing on learning and movement in order to improve body movement and function (The Feldenkrais Guild UK).

Magnetic resonance imaging (MRI): the magnetic properties of hydrogen and its interaction with magnetic fields and radio waves are used to produce high resolution images of the lumbar body.

Interval for x, *a* and *b*: range of values between lower and upper limits. Can be of different types as follows: from *a* included to *b* included ($a \leq x \leq b$), from *a* excluded to *b* included ($a < x \leq b$), from *a* included to *b* excluded ($a \leq x < b$), from *a* excluded) to *b* excluded ($a < x < b$) (Montgomery and Runger, 2011).

Box plot: graphical display (Figure 1.1-1) of data showing five statistics: minimum, first quartile, median, third quartile, and maximum (Montgomery and Runger, 2011).

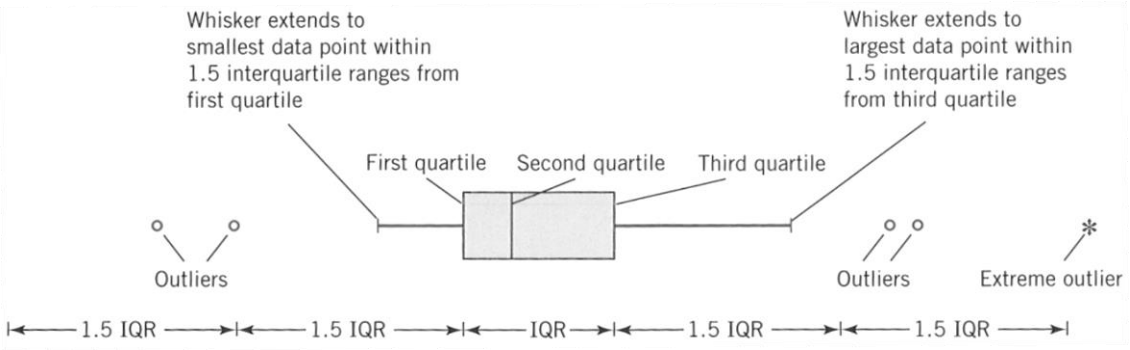

Figure 1.5-1 Description of a boxplot. Edited from Montgomery and Runger (2011). Interquartile range (IQR).

Relative percentage change or difference between x and y $= \left(\frac{y-x}{|x|}\right) 100$ (Montgomery and Runger, 2011)

Reliable: that yields the same or comparable result in different statistical trials (Montgomery and Runger, 2011).

Accurate: that provides a correct measurement or result (Montgomery and Runger, 2011).

CHAPTER 2 Literature Review

2.1. Introduction

In the first section of this review, the biomechanics of the human spine is briefly outlined. The effect of osteoporosis on the biomechanics of the spine is highlighted. In the next section, the current management of the osteoporotic spine is presented briefly. Physical activity as a non-pharmacological treatment is highlighted and compared with whole body vibration. In the third section, current knowledge on the effects of vibration on the spine is explained based on in vitro and in vivo studies. The last section explores current techniques used to assess the mechanical properties of the healthy and osteoporotic spine in vivo.

2.1.1. Spine biomechanics

The human spine consists of a series of vertebrae interconnected by intervertebral discs. The curvature of the spine may increase the strength of the overall structure and help the body balance and absorb shocks during gait (Tortora and Grabowski, 2003). Intervertebral discs absorb vertical shocks (Arakal et al., 2011) and distribute loads evenly to the adjacent vertebrae (Adams and Dolan, 2005). Vertebrae vary in size and shape according to the region of the spine they belong to, for instance lumbar vertebrae are the largest and strongest (Kolta et al., 2012). Thoracic vertebrae also articulate with the ribs while the fifth lumbar vertebra articulates with the first sacral vertebra (Tortora and Grabowski, 2003). Spine motion can be studied as a single structure between the pelvis and the head as well as divided into sections. An intervertebral joint has six degrees of freedom: three translations or displacements and three rotations (Wu et al., 2002). A standard joint coordinate system was recommended by Wu et al. (2002) for the study of the biomechanics of the spine (Figure 2.1-1). The main components are the origin (o), a vertical axis (y,Y) pointing to the head, a mediolateral axis (z,Z) pointing right and an anteroposterior axis (x,X) pointing to the anterior direction. From these axes, displacements and rotations can be defined as flexion or extension (e1) with mediolateral translation, axial rotation (e3) with proximodistal translation and lateral bending (e2) with anteroposterior

translation. The articulation between vertebrae allow for flexion, extension, lateral bending and rotation.

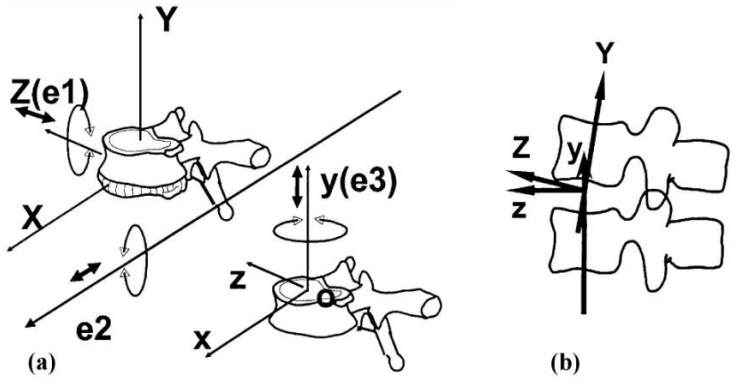

Figure 2.1-1 Joint coordinate system for the spine. (a) proximal (XYZ) and distal (xyz) JCS. (b) Interaction of coordinate systems, adapted from Wu et al. (2002)

Structural behaviour of the spine is dependent on its morphology, geometry and material properties. Material properties are independent of structure or geometry (Davison et al., 2006). The biomechanical properties of human lumbar and thoracic spines have been widely studied in-vitro (mainly) as well as in vivo (Wilke et al., 1999, Takahashi et al., 2006, Schultz et al., 1982). In vivo studies have looked mainly at load and pressure of lumbar intervertebral discs (Schultz et al., 1982). In vitro studies have looked at multiple levels of the spine and have provided detailed data on stiffness (Busscher et al., 2009, Gardner-Morse and Stokes, 2004), shear loading (Skrzypiec et al., 2012), range of motion (Panjabi et al., 1976, Deitz et al., 2011) and load-displacement curves (Panjabi, 1976). In most of these studies, segments of human cadaveric spines were examined with their muscle and other soft tissue removed (Panjabi, 1976, Gardner-Morse and Stokes, 2004, Busscher et al., 2009). An additional disadvantage of in vitro biomechanical tests of the spine or any of its elements is that the natural alignment of the spine and its segments is often not considered. This alignment has significant effects of the compressive failure limits of lumbar and thoracic spines (Campbell-Kyureghyan et al., 2011).

2.1.2. Ageing spine and osteoporosis

Ageing is associated with tissue degeneration and different tissues (cartilage, bone, ligaments, muscles) are affected in various ways. Here degeneration implies both mechanical or structural changes as well as changes at cellular level like nutrition and composition (Adams and Dolan, 2005). For example, one major biochemical change of cartilage is the loss of proteoglycans leading to a stiffer tissue. Degeneration of intervertebral discs is common and they often collapse (Adams et al., 1996, Stokes and Latridis, 2004). In vitro tests by Twomey and Taylor (1985) confirmed that the loss of transverse trabeculae in lumbar vertebrae due to age contributes to a change of shape of both vertebrae and intervertebral discs. Degeneration of spinal joints as well as ligaments losing material properties lead to laxity and instability (Araghi and Ohnmeiss, 2011). Vertebral size and shape, as indicators of geometric dimensions, increase with ageing in the lumbar spine but not in the thoracic spine (Kolta et al., 2012). However, it is not known if this is a natural response for bone loss or if it is a determinant for higher risk of fracture at the lumbar spine (Kolta et al., 2012). Changes in posture with age affect important muscles that stabilize the spine during standing and sitting postures, leading to spinal deformity due to stress redistribution, altered motion and pain (Adams et al., 1996). Added to this, in-vitro tests by Taylor and Twomey (1986) showed that the articular cartilage and subchondral bone of the anterior coronally oriented third section of the zygapophyseal joint of lumbar vertebrae is subjected to changes that are related to loading during flexion throughout life. Taylor and Twomey (1986) also observed that the subchondral bone plate retains its shape through life regardless of osteoporosis (Taylor and Twomey, 1986). Because the spinal elements are interconnected, the degeneration and injury at one level usually causes adjacent levels to degenerate and become vulnerable to injury. Normal forces generated during gait can injure abnormally weak bone and tissue (Adams and Dolan, 2005). Other factors that further escalate degeneration with age are malnutrition, smoking, obesity, lack of exercise, osteoarthritis, spinal deformities and osteoporosis (Mok et al., 2011).

Natural bone loss or demineralization occurs after approximately the age of 35 (WHO, 1994). In old age, bone resorption naturally outpaces bone deposition (Figure 2.1-2). But this bone remodelling

process may be severely augmented due to other factors such as a decline of oestrogen production in women, low peak bone mass at a younger age, family history of low bone mass, alcohol and tobacco consumption, lack of exercise and also low body weight. This process of demineralization that causes bones to become fragile and brittle is called osteoporosis (Bouxsein, 2005). Trabecular bone is affected more than cortical bone therefore the risk of fracture is greater in bones with predominant trabecular bone like vertebrae, distal radius and head of the femur (Mc Donnell et al., 2007). In biomechanical terms, spine fractures may occur while performing normal daily life activities (Silva, 2007, Tortora and Grabowski, 2003). Because the risk of fracture of an osteoporotic spine is high, the risk of falling becomes important (WHO, 1994, Bouxsein, 2006). Hence, biomechanical studies focus on the study of risk of falling (Unnanuntana et al., 2011) with less emphasis on the structural and material properties of the osteoporotic spine as a system (Adams and Dolan, 2005).

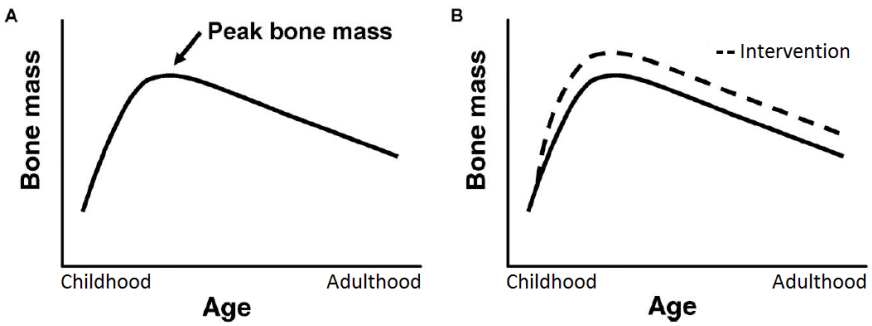

Figure 2.1-2 Conceptual graphs of bone mass as a function of age for a normal person (A) and for an example of how an assumed intervention during childhood will have a continous effect on bone mass thoughout life (B). Edited from (Gafni and Baron, 2007)

Mainly BMD is used in clinical practice for the diagnosis, treatment and monitoring of the osteoporotic spine. There is no clinical test to determine the structural and material properties of the osteoporotic spine in vivo other than its mineral content (Unnanuntana et al., 2011). Healthy and osteoporotic femoral and cortical bone have been widely tested in vitro and all agree that the mechanical properties of bone deteriorate with age and with osteoporosis, specifically with reduced stiffness (Burstein et al., 1976, Dickenson et al., 1981). Osteoporotic spinal units have also been tested

in-vitro. For instance, Skrzypiec et al. (2012) found that BMD correlated positively with peak shear force of L2-L3 human lumbar segments. One drawback of in vitro studies is that the determination of BMD can be underestimated (Skrzypiec et al., 2012). Another disadvantage is that the degrees of freedom of the functional spinal unit are reduced during testing. These studies are also limited to the cadaveric specimens therefore there is scope for biomechanical research that would consider the mechanical and structural properties of the spine in vivo (Bouxsein, 2005, Mc Donnell et al., 2007).

Spinal range of motion (ROM) and velocity decrease with osteopenia and osteoporosis (Tsauo et al., 2002). In contrast, Yang et al. (2009) found that neither osteoporosis nor osteopenia appear to affect lumbar spinal mobility. However, Yang et al. (2009) found that a height increment in the middle of lumbar intervertebral discs (expansion) was associated with osteoporosis. Yang et al. (2009) also found that the anterior height of lumbar vertebrae was slightly larger in osteoporotic subjects when compared with healthy ones. Yet these studies focus on flexion and extension tasks and not in daily life physical activities, thus it is unknown how structural changes due to osteoporosis affect spinal biomechanics during tasks such as walking, turning, stair negotiation or during exercise. It has been suggested that people uses a relatively small percentage (4% to 59%) of their full functional range of ROM during daily life physical activities. Interestingly, personal hygiene activities (hand and hair washing, shaving, make up application) require a similar ROM compared with walking and stair negotiation. Studying the clinical implications of walking and stair negotiation may provide information on a broader range of daily physical activities than just these activities alone.

2.1.3. Summary

The human spine is a complex structure, with regions defined by function, geometry as well as tissue composition. Biomechanical analysis of the spine has favoured mainly the investigation of material properties such as bone and cartilage. Biomechanical analysis has investigated functional spinal units in-vitro, restricting in this way the real degrees of freedom and potentially biasing results by removing muscles and other soft tissue surrounding the spine in real life. In general, mobility, stability and load bearing capacity are possible due to the multifactorial nature of the spine elements.

In vivo biomechanical studies of the spine have the scope for further non-invasive research that could analyse the spine as a system of complex structures and during daily life physical activities. It is necessary to understand the effect of osteoporosis on the structural properties of the human spine in vivo. The current evaluation of the spine in vivo through a BMD measurement and the approximation of the risk of falling do not fully account for the structural changes due to osteoporosis, soft tissue properties and for the effect of daily life activities.

2.2. Management of the osteoporotic spine

In this section, current pharmacological and non pharmacological interventions for osteopenia and osteoporosis will be presented. Non pharmacological treatments such as physical activity and whole body vibration will be outlined in more detail to establish a clear difference between vibration produced by a machine and that produced during physical activity.

2.2.1. Pharmacological and surgical interventions

Current pharmacological treatments available are focused on prevention and treatment of osteoporosis mainly for postmenopausal women. These treatments decrease risk of fracture by increasing BMD. BMD increment is achieved either by modifying biochemical paths of cells in charge of bone remodelling or by simply increasing calcium content in the blood and its retention in the body. There are currently five treatments from which medical doctors can choose: bisphosphonates, parathyroid hormone, oestrogen therapy, calcitonin, calcium and vitamin D supplements. It is not in the scope of this thesis to explain in detail how each pharmacological treatment works. However it is important to note that every pharmacological therapy involves adverse effects and does not guarantee a full recovery of bone health. Instead, spinal BMD is only increased from 1% to 15% and vertebral fracture risk reduced from 30% to 83% depending on the gender of the subject and drug and time used (Burr et al., 2002, NOF, 2010, The DIPART Group, 2010, Amrein et al., 2012). This suggests that a pharmacological approach for the treatment of osteoporosis is not sufficiently efficient alone and that other non pharmacological interventions need to be applied. It is also important to note that pharmacological treatments are focused on treatment of osteoporosis rather than its prevention. Most drugs have been clinically tested to treat severe osteoporosis but not to prevent it. Similarly, long term

use effects are not yet known (Burr et al., 2002, NOF, 2010, The DIPART Group, 2010, Amrein et al., 2012).

Surgical intervention is often used when vertebral fractures due to osteoporosis lead to other complications such as pain, disability, deformity, neurologic deterioration and damage to other organs which the spine is no longer able to protect (Gebauer and Khanna, 2010). Two minimally invasive methods are currently employed to provide strength to the fractured vertebra due to osteoporosis through the injection of bone cement (methylmethacrylate or PMMA). These are vertebroplasty and kyphoplasty (Gardner, 2011, Truumees, 2002, Heini, 2011). Although these surgical procedures still need further investigation to prevent adjacent level fractures, they are capable of restoring vertebral stiffness and enable the restoration of load transfer near normal values (Luo et al., 2010a, Luo et al., 2010b). These surgical procedures are influenced by BMD, fracture severity and disc degeneration (Luo et al., 2007).

2.2.2. Diet

Good nutrition throughout life is one of the factors that can be modified to prevent osteoporosis in later life (IOF, 2012). 98% of the human skeleton is composed of calcium (Tortora and Grabowski, 2003). Dietary calcium and vitamin D as well as its supplementation maintain bone mass and reduce bone loss but their benefits have been questioned since these are not the only components that maintain bone (Burr et al., 2002, Zhu and Prince, 2012). The beneficial effects of calcium have been studied mainly for reducing the risk of hip fracture. Other dietary components that affect bone health are protein, phosphorus, zinc, magnesium, iron, vitamin K, vitamin B12, vitamin A, fatty acids, phytoestrogens, caffeine, tobacco and alcohol (Burr et al., 2002, Kerstetter et al., 2007, Martínez-Ramírez et al., 2012, Unnanuntana et al., 2011, Higgs and Kessenich, 2010) (Rapuri et al., 2007). (NOF, 2010, Berg et al., 2008, Eleftheriou et al., 2013). (NOF, 2010, Eleftheriou et al., 2013). A varied diet with adequate servings from each food group would provide the right energy and components to maintain healthy bones, while sunbathing and while avoiding excessive consumption of damaging components such as caffeine, tobacco and alcohol. Maintenance of adequate nutrition is a treatment used in clinical practice along pharmacological treatments to reduce fracture risk.

2.2.3. Whole body vibration therapy

The effects of vibration on the body have been studied for the treatment of sarcopenia and osteoporosis. Vibration may amplify bone mechanotransduction signals which in turn induce bone cells to respond by adapting its structure and mineral content (Qin et al., 1998, Qin et al., 2002, Judex and Rubin, 2010, Totosy de Zepetnek et al., 2009, Ozcivici et al., 2010, Letechipia et al., 2010, Chen et al., 2010). This means that bone will respond to vibration by becoming stronger.

Low bone mass due to the disuse of the skeletal system (astronauts and paraplegics) or due to disease (osteoporosis and osteopenia) has been targeted through whole body vibration (WBV) in order to reduce bone loss. WBV consists in exposing the human body to forced sinusoidal oscillations created by a motor below a platform. WBV is transmitted to a person standing on the platform (Rittweger, 2010). Six determinants have been identified for the description of WBV therapy: shape (sinusoidal), direction (vertical or rotational), modality (synchronous, side alternating and tri planar), frequency (Hz), magnitude (acceleration in m/s^2or g), displacement (mm), body posture (for example standing or sitting) and its duration (Wysocki et al., 2011). There is wide variability in the parameters offered by commercial vibration platforms (Prisby et al., 2008, Kasturi and Adler, 2011, Wysocki et al., 2011). Vibration is commonly applied in the vertical direction, ranging from 0.7 to 12 mm at 12 to 90 Hz with a magnitude of 0.3 g to 10 g at intermittent cycles, every 15 seconds to every 30 minutes during a standing position (with or without knee flexion and even while performing some exercises) from once a week to daily for 8 to 72 weeks (Judex and Rubin, 2010, Kasturi and Adler, 2011, Slatkovska et al., 2011, von Stengel et al., 2011a, Wysocki et al., 2011). Platforms that provide acceleration of less than 1 g (low intensity or magnitude) are used for the treatment of osteoporosis, however research studies have stimulated at greater than 1g, which in theory should be used in high intensity vibration exercise for healthy populations only (von Stengel et al., 2011a). However, no organisation provides accreditation or training for the use of WBV platforms in clinical professional settings. Training exclusively on the use of the equipment is seldom provided, no training in the clinical application for prevention and treatment of osteoporosis is available (Wysocki et al., 2011). In a review by the International Society of Musculoskeletal and Neuronal Interactions (ISMNI), members aimed to

provide a standard suggestion on how WBV intervention studies should report vibration characteristics (Rauch et al., 2010). Rauch et al. (2010) explain that there are variables that are not normally reported but which have some effect on the acceleration delivered to subjects. These variables are: the rigidity of the platform plate, foot position and type of footwear. Rauch et al. (2010) also stressed the importance of characterizing the vibration amplitude and frequency delivered by each vibration device since there is no entity to regulate vibration plates commercialy available (Rauch et al., 2010). The ACSM in an attempt to provide research based guidelines in the use of vibrating force plates for the treatment of osteopenia and osteoporosis, published a list of recommendations (Tomás et al., 2011). First, individuals with an osteoporotic vertebral fracture should avoid WBV. Secondly, in order to increase BMD, WBV should be vertical from 35 to 40 Hz with displacements between 1.7 to 2.5 mm of magnitude lower than 1 g, in a session of no longer than 30 minutes with intermitent exposure of no longer than 30 seconds each. Third, squats, deep squats and lunges as a form of exercise during the session are permitted. And lastly, this should take place at least three times a week. Yet, the ACSM fails to give a detailed explanation of how this recommendation has been reached. Interestingly, Tomás et al. (2011) also indicate that during the exercise over the platform, knees should not be locked in order to avoid transmission of vibration to the trunk, which suggests that the ACSM considers that WBV is not for the 'whole body' but just for the lower limbs. An individual's posture alter the transmissibility of vibration to the head and chest, for example knee flexion (Rittweger, 2010). Thus this is a reason to consider that WBV may not safe to stimulate the osteopenic and osteoporotic spine.

Several contraindications exist, mainly derived from research on vibration during occupational and transport tasks (Rubin, 2006). In general, WBV is contraindicated for people with kidney or bladder stones, arrhythmia, pregnancy, epilepsy, seizures, cancer, untreated orthostatic hypotension, people with any type of implants, recent surgery of any kind, thrombosis, rheumatoid arthritis, cardiovascular disease, severe diabetes and migraine (Rubin, 2006).

Research evidence has contrasting results regarding the effects of WBV on spinal BMD while employing inconsistent terminology and widely different WBV interventions (Lorenzen et al., 2010, Hill et al., 2009, Judex and Rubin, 2010, Mikhael et al., 2010, Liu et al., 2011, Totosy de Zepetnek et al., 2009, Wysocki et al., 2011, Slatkovska et al., 2011). In a double blind, placebo controlled pilot study on the effect of WBV, postmenopausal women with a weight lower than 65 kg achieved a 3.4% increment in spinal BMD (Rubin et al., 2004). In contrast, Slatkovska et al. (2011) concluded that WBV had no effect on areal BMD of the lumbar spine of postmenopausal women when treated for 12 months with 0.3 g at 30 Hz and 90 Hz 20 min daily. These ranges of vibration characteristics are so wide due to the fact that research studies have employed both low and high magnitude vibration which had different purposes (the first for the treatment of osteoporosis and the second for exercise training in healthy subjects). Moreover participants subjected to WBV report side effects such as dizzines, nausea and chronic shin and foot pain. Overall, this study indicates that WBV is not an effective therapy for preventing bone loss, for increasing mineral density and for improving bone structure of osteopenic and osteoporotic women on calcium and vitamin D supplementation. According to a review by Cheung and Giangregorio (2012) literature shows that low magnitude, high frequency WBV does not improve BMD and bone structure in postmenopausal women.

Gómez-Cabello et al. (2012) favour WBV over physical activity since they suggested that both show modest improvements of BMD and bone structure while WBV is a good alternative for individuals intolerant to exercise (perhaps due to pain) and those with limited mobility. Yet this recommendation should be well delimited in order to prevent people that would potentially benefit from progressive strength training and other health benefits brought by physical activity, rather than opting for WBV for which long term efficacy and side effects are still unknown. Wysocki et al. (2011) highlighted several reasons for considering WBV therapy as an intervention that cannot be considered safe, reliable and effective. It is not approved by the Food and Drug Administration (FDA) of the USA, the design and protocols of WBV platforms are not standardized and it has not been established if it leads to significantly important preservation or increment of BMD nor its effects in bone structure in

humans. Finally, WBV does not enhance the effect of 18 months of exercise on lumbar BMD in postmenopausal women (von Stengel et al., 2011b).

In summary, there is more evidence pointing out the benefits of physical activity than WBV. Beneficial effects of WBV in vivo for the spine have not been demonstrated in contrast with small benefits seen when performing physical activity. In general, the efficacy and safety of WBV is unknown. Vibration protocols vary considerably. Most studies agree that more research is needed in order to determine if WBV should be excluded completely from available treatments of osteoporosis or if it offers any advantage or contribution when mixed with other treatments.

2.2.4. Physical activity and exercise

Physical activity recommendations for the treatment of osteoporosis are dependant of the way health institutions and organizations classify physical activity. The American College of Sports Medicine (ACSM) has provided definitions for physical activity and exercise (Chodzko-Zajko et al., 2009). Physical activity is any body movement produced that requires muscular contraction and energy expenditure. Exercise is a planned, structured and repetitive movement that has the objective of improving or maintaining physical fitness. Similarly, aerobic exercise training refers to sustained rhythmic muscular movements while resistance exercise training refers to when muscles hold against a weight. Flexibility exercise consists in preserving or extending the range of motion of a joint while balance training consists in increasing the strength of the lower body towards reducing the risk of falling (Chodzko-Zajko et al., 2009).

Overviews of the multiple classifications that are currently used are presented in Table 2.2-1. The British Heart Foundation National Centre (BHFNC) classifies physical activity in three types according to expenditure of calories: everyday, active recreation and sport. Physical activity has also been classified in three types according to its intensity: moderate, vigorous and muscle strengthening (UK Chief Medical Officers, 2012). In comparison, Langsetmo et al. (2012) appear to exclude normal walking from any classification but include brisk walking under moderate activity. Langsetmo et al. (2012) also classified activity according to expenditure of calories but defined only two categories:

moderate and vigorous (Table 2.2-1). The American College of Sports Medicine (ACSM) classifies aerobic exercise by intensity in only two categories while classifying physical activity according to specificity as well (Table 2.2-1). The World Health Organization (WHO, 2012) classifies physical activity according to intensity into moderate and vigorous but gives more examples in contrast with the ACSM, BHFNC and Langsetmo et al. (2012). The WHO adds that the classification suggested may change between individuals, which suggests that an additional way to objectively measure physical activity is needed (Table 2.2-1).

Older adults often do not benefit from prescribed physical activity due to "vague or inappropriate instructions" (Mc Dermott and Mernitz, 2006). Not only clinicians and patients are affected by the lack of guidelines for the prescription of physical activity but also research studies which reach conclusions and recommendations based on classification of physical activity that often overlap with other studies (Hurley and Armstrong, 2012). An effective exercise prescription would ideally include frequency, intensity, type, time and progression according to individual capabilities and current health status while addressing expectations and barriers, but most importantly it should provide a way to reliably measure physical activity (Mc Dermott and Mernitz, 2006). Skerry (2008) pointed out that identical mechanical environments can be perceived differently between subjects due to the fact that bone response depends on many factors. This means that the same physical activity can be perceived as an overload stimulus, as a habitual load or as a disuse situation, depending on individual capabilities. A recent study by Gába et al. (2012) evaluated the feasibility of objectively measuring physical activity through accelerometers on the hip. No correlation was found between physical levels expressed in intensity levels (METs) and femoral BMD (Gába et al., 2012). Gába et al. (2012) also concluded that body composition may play an important role in determining BMD. However, it is possible that interpreting the vertical accelerations produced during daily life physical activities as METs is not the optimal way since it does not offer any link with bone health. Perhaps vertical acceleration has to be correlated as it is but with structural indicators (transmissibility, peripheral quantitative computed tomography (pQCT), magnetic resonance imaging (MRI)) rather than BMD.

Thus, it is evident that there is the need to find a technique to measure physical activity across individuals with different capabilities and health.

Table 2.2-1 Classifications of physical activity

Institution or Organization	Classification				
According to expenditure of calories (UK Chief Medical Officers, 2012) for the BHFNC	**Everyday Activity**	**Active Recreation**	**Sport**		
	Active travel (walking or cycling), heavy housework, gardening, occupational activity, DIY	Recreational walking, recreational cycling, dancing, active play	Sport walking, regular cycling, swimming, exercise and fitness training, competitive activity, individual pursuits, informal sport		
According to intensity (UK Chief Medical Officers, 2012) for the BHFNC	**Moderate**	**Vigorous**	**Muscle strength**		
	Brisk walking, bike riding, dancing, swimming, active travel	Running, playing sport, aerobic exercise, use cardiovascular gym equipment	Weight training, work with resistance bands, carrying heavy loads, heavy gardening, push up, sit up		
According to intensity (Langsetmo et al., 2012)	Brisk walking, bike riding on level ground, housework, golf, bowling	Weight lifting, moving furniture, loading and unloading trucks, shovelling, manual labour	-		
Aerobic exercise according to intensity (Winter-Stone, 2005) for the ACSM	Brisk walking, brisk hiking on level ground, low impact aerobics, moderate pacing dancing (waltz), tennis (doubles), light rowing	Running or jogging, fast or race walking, fast paced dancing (salsa, jitterbug, disco), high impact aerobics, tennis (singles), fast rowing	-		
According to intensity by the World Health Organization (WHO, 2012)	Brisk walking, dancing, housework, gardening, active involvement in sports with children, walking domestic animals, building tasks, carrying or moving objects (<20 kg)	Running, walking up a hill, fast cycling, aerobics, fast swimming, competitive sports, heavy shovelling or digging, carrying heavy loads (>20 kg)	-		
According to specificity (Winter-Stone, 2005) for the ACSM	**Aerobic**	**Resistance**	**Flexibility**	**Balance**	**Impact**
	Walking, jogging, swimming, cycling, dancing, cross country skiing, hiking, rowing, marching in place	Weight lifting	Stretching, yoga, Tai Chi	Yoga, Tai Chi	Jumping, hopping, skipping integrated to: aerobic dancing, stepping routines, basketball, volleyball, gymnastics

2.2.4.1. Exercise for healthy older adults

Physical activity has been considered as a means of maintaining good health in general, not only bone health, for individuals of all ages (Cech, 2012, Chodzko-Zajko et al., 2009). The ability of exercise to make bones stronger, maintain current BMD and promote a higher BMD has been studied through bone scans such as dual X-ray absorptiometry (DXA), QCT and MRI mainly in healthy adults and children (Hamilton et al., 2010b, Cousins et al., 2010, Bailey et al., 2010). These studies have focused more in long bones (tibiae, radius, femur) rather that in the spine. Exercise promotes large improvements in bone strength but small gains in BMD (Hamilton et al., 2010b, Cousins et al., 2010, Bailey et al., 2010). Numerous clinical studies have provided evidence that strenuous physical activity significantly affects bone mass in children while in young and older adults it only produces a modest increment in bone mass (IOF, 2012, UK Chief Medical Officers, 2012). Langsetmo et al. (2012) found an association between spinal BMD improvement and physical activity performed over a period of five years, but this effect was only observed in men. The main disadvantage of this study is that it obtained information on the physical activity performed by volunteers through self-reported questionnaires. Mayer et al. (2011) concluded through a review that strength training at medium and high intensities, several times per week for months would reduce bone loss in older adults under instruction, giving light to the fact that dosage has not been outlined conclusively in the literature. The BHFNC has published a report on physical activity recommended for healthy adults (19-64 years old) and older adults (65+ years old) in the United Kingdom, in order to reduce the risk of musculoskeletal conditions (UK Chief Medical Officers, 2012). For both groups 150 minutes of moderate and 75 minutes of vigorous activities are recommended. Only if older adults are already considered as 'active' can engage in vigorous activities. The ACSM recommends that healthy older adults should engage in aerobic exercise, muscle strengthening and flexibility exercises to limit the development and progression of chronic diseases that characterize human ageing (Chodzko-Zajko et al., 2009).

The effect of exercise on bone is specific to the region that is being stimulated with a greater load than usual as well as the level of skeletal maturity (Kohrt et al., 2004). If stimulation is removed

BMD decreases (Kohrt et al., 2004). Type, intensity, frequency, duration and rest between exercise sessions determine the outcome (Kohrt et al., 2004, Manske et al., 2009). Dynamic loading is more effective than static and achievement of bone strain is more important than frequency of loading (Manske et al., 2009). In addition, gravitational and muscular forces may have an important role in determining bone mass and structure (Kohrt et al., 2009).

There is a general view that at an older age load bearing intensity should be diminished in order to prevent injury (Mayer et al., 2011), however this is not supported by research evidence. Current reviews in physical activity for healthy older adults agree that high intensity loading is not only endured by healthy older adults but also necessary to sustain general health (Chodzko-Zajko et al., 2009).

2.2.4.2. Exercise for adults with osteoporosis

Exercise has also been classified for prevention of osteopenia and osteoporosis, for conservation of current BMD and for restoration of BMD during osteopenia and osteoporosis (Chodzko-Zajko et al., 2009). Maximizing peak bone mass by the time individuals reach the age of 30-35 years old is currently considered as a preventive measure for osteopenia and osteoporosis (Cech, 2012). After the age of 35 to 40 years old, bone mass decreases approximately 0.5% per year (Chodzko-Zajko et al., 2009).

It has been assumed that exercise has the same beneficial properties for bone in older adults with osteoporosis and osteopenia as it does for healthy adults. In contrast, it has been previously suggested that physical activity may not be beneficial for people with osteoporosis but rather cause fractures (Rittweger, 2006). Thus, the greatest challenge is to determine the characteristics of physical activity that individuals with risk of fracture should perform. Exercise prescription guidelines for osteoporotic individuals are available from the American College of Sports Medicine (Winter-Stone, 2005), International Osteoporosis Foundation (IOF, 2012), from the World Health Organization (WHO, 2003), National Osteoporosis Society from the United Kingdom (NOS, 2012), a clinical practice guideline (NOF, 2010) and from numerous reviews (Martyn-St James and

Carroll, 2009, Martyn-St James and Carroll, 2008, Going and Laudermilk, 2009, Schmitt et al., 2009, Cheung and Giangregorio, 2012, Gremeaux et al., 2012, Howe et al., 2011, Marques et al., 2011, Gómez-Cabello et al., 2012, Lips and Van Schoor, 2006). Books by experts in physical education have also compiled research papers and combined them with their experience with people with osteoporosis (Daniels, 2008, Van Norman, 2010). There is a lack of evidence based research of the threshold at which physical activity would either improve bone health or increase the risk of fractures in already osteoporotic bone (Rittweger, 2006, Hamilton et al., 2010a, Gremeaux et al., 2012, Kohrt et al., 2004, Cheung and Giangregorio, 2012, Mayer et al., 2011). This suggests that there is the need to provide a standard way to measure physical activity.

The position of the WHO has been that exercise should be encouraged throughout life while randomized controlled trials are necessary to provide evidence on the effectiveness of exercise for fracture prevention (WHO, 2003). The WHO also notes that while exercise should be encouraged it is not the only treatment available for osteoporosis and as such other interventions should complement treatment.

Kohrt et al. (2004) noted that there was no technique to quantify exercise intensity in terms of bone loading, only conventional methods such as metabolic and cardiovascular stresses were used (percent maximal heart rate, percent of one repetition maximum, maximal oxygen consumption). Kohrt et al. (2004) published physical activity recommendations for the ACSM which consisted in not only maintaining an active life style but to include weight bearing endurance and resistance activities as well as activities to improve balance and prevent falls. Kohrt et al. (2004) also noted that there are opposing results due to the overlap in the type of activity tested based on its objective (to promote strength, endurance, balance or flexibility) as well as the duration and frequency of the physical activity tested. The ACSM indicates that activities such as swimming and cycling do not strengthen bone since these are not weight bearing exercises (Winter-Stone, 2005). Physical activities identified to improve bone health are: aerobic exercise that promotes weight bearing (fast walking, jogging, running, stair stepping and aerobic dance), weight lifting and jumping. However,

jumping is generally recommended to increase bone mass at the hip not the spine. The types of aerobic exercise that seem to not affect bone mass are slow and moderate walking and aerobic exercise in water. Martyn-St James and Carrol (2008) also found that walking from 6 to 12 months does not preserve BMD in the lumbar spine of postmenopausal women. Therefore this was a major indicator that other forms of exercise capable of maintaining or increasing BMD have to be identified. Winter-Stone (2005) suggests that people with osteopenia can safely do exercise at moderate to high intensity in order to improve bone health. While people with osteoporosis should perform exercise under supervision of a health care team which should judge the appropriate amount of exercise to perform only after determining the baseline fitness individually (Winter-Stone, 2005). The types of exercise that people with osteoporosis should avoid according to the ACSM and Winter-Stone (2005) are: very high impact exercise (jumping with added weight), back flexion against resistance, overhead weight lifting, lifting a weight well away from the body and fast twisting (golf). The recommendation of the ACSM is to perform exercise even with osteopenia and osteoporosis (Winter-Stone, 2005).

The IOF recommends physical activity to target mainly posture, balance, gait coordination and hip and trunk stabilization instead of aerobic fitness. This is based on a study performed exclusively on Canadians in the year 2002. Furthermore, it stresses that dynamic abdominal exercises (sit ups, trunk flexion) and high impact exercises are contraindicated for people with osteoporosis due to the risk of vertebral fractures. These last contraindicated exercises are made based on a publication from 1984 (IOF, 2012) which may not be reliable since there are currently new approaches that recommend progressive weight bearing training. The American National Osteoporosis Foundation recommends regular weight bearing exercise but low impact for people with osteoporosis, similarly, it recommends fast walking, low impact aerobics, Tai-Chi, yoga, stair climbing, dancing and tennis (NOF, 2010, NOF, 2012). It stresses that before an individual with osteoporosis can initiate any of these activities the opinion of a doctor is required, but fails to direct to research based evidence on how each of these activities would benefit or put the individual in higher risk of fracture.

The National Osteoporosis Society (NOS) of the UK agrees with the ACSM that swimming and cycling are not weight bearing activities therefore these do not have any effect on bone mineral density (NOS, 2012). Exercise recommendations for people with spinal osteopenia or osteoporosis with no previous fractures are mixed weight bearing exercises, jogging and stair climbing. Exercise recommendations for people with osteopenia or osteoporosis who have experienced a fracture are strength training using body weight as resistance, weight bearing aerobic activities (walking, dancing, low impact aerobics) as well as exercises for flexibility and stability. Exercises that should be avoided with high risk of fracture are high impact fast moving exercises (jumping, running, jogging, skipping, horse riding and skiing), forward bending, twisting and sit ups (golf, tennis, bowling and some yoga poses). The NOS stresses the importance of understanding that there is no exercise prescription that will "fit all", instead people with osteoporosis can start with safe physical activity and increase difficulty with caution (NOS, 2012). Yet there is no way to measure if the weight bearing exercises chosen between a patient and the exercise professional are indeed stimulating the spine. It is also worrying that the NOS recommends exercise for which there is no evidence supporting its benefits on the osteopenic and osteoporotic bone, less its safety. These exercises are trampolining, walking poles (Nordic walking and pacer poles), netball, basketball, football, and hockey as well as racket sports such as squash, tennis and badminton. Similarly the NOS recommends line and Irish dancing as well as the use of weighted vests without reliable research evidence on the effect of these on the osteoporotic spine (NOS, 2012). The BHFNC has published a report on physical activity recommended for 'older adults at risk of falling' (65+ years old) in the United Kingdom, in order to reduce the risk of musculoskeletal conditions. The BHFNC also recommends physical activity to improve balance and coordination (two days a week) and physical activity to improve muscle strength (two days a week). The limitation of this guideline is that it recommends these physical activities as a way of maintaining health but does not mention any particular activity that an older adult with osteoporosis could safely perform in order to either stop bone loss or to increase it. However, it does mention that these guidelines need to be individually adjusted based on exercise capacity and special health risks. Nevertheless, it fails to

provide guidance on how to determine that exercise capacity of a person with fracture risk for example.

A review by Lips and Van Schoor (2006) concluded that walking and moderate strength training was the best recommendation of type of exercise for patients with osteoporosis but noted that there was no evidence that exercise can decrease the number of fractures, instead most literature focus in fall prevention. As part of a research agenda, Lips and Van Schoor (2006) recommended to find the type of exercise that is most effective to prevent fractures and falls. Mc Dermott and Mernitz (2006) indicated that an exercise prescription for "frail" older adults should be chair and bed based as a starting point. Martyn-St James and Carrol (2009) found that mixed loading exercise programmes may reduce spinal bone loss in postmenopausal women. Martyn-St James and Carroll (2009) recommended exercise such as jogging mixed with walking and stair climbing and exercise programmes that combine impact and resistance exercise. Going and Laudermilk (2009) extrapolated animal studies to exercise programmes for humans. Going and Laudermilk (2009) found that an effective exercise programme should be dynamic, exceed a threshold intensity and frequency and be brief and intermittent while promoting load patterns to which the bone is not accustomed. The only activity that could fulfil these requirements was resistance exercise, which was considered as safe for older adults with osteoporosis given that squats and curl ups are excluded from any exercise programme. Besides this, Going and Laudermilk (2009) acknowledged that non mechanical factors can modify the human response to exercise, making it difficult to interpret studies looking at specific exercise programmes and populations. Schmitt et al. (2009) concluded that the intensity of physical activity determines its effectiveness in the prevention and treatment of osteoporosis but the risk of injury should not be neglected when recommending exercise for the older population. The authors point out correctly these considerations to be taken and also that good quality long term studies are needed to identify optimal physical activities outside the existing programmes (aerobics, Tai Chi and walking) that seem less effective in the treatment of osteoporosis. Marques et al. (2011) concluded that exercise protocols that combine impact activity with high magnitude resistance training were most effective in achieving a small

increment of BMD at the lumbar spine of up to 0.011 g/cm^2. Marques et al. (2011) recommended studies to exclude low intensity impact exercise and concentrate on a combination of high odd impact loading, muscle strength and balance. Yet it is not clear how excluding low impact activities would affect the maintenance of current BMD in the osteoporotic spine. Moreover, Marques et al. (2011) highlighted that studies need to include osteoporotic subjects in order to reliably establish the effect of exercise. Similarly, Marques et al. (2011) indicated that further research must include biomechanical parameters related not only with BMD but with size and structure. A review for the Cochrane Collaboration (Howe et al., 2011) found that postmenopausal women that performed a combination of exercises had on average 3.2% less bone loss than those who did not exercise. The types of exercise that studies tested on the effect on BMD included static weight bearing, dynamic weight bearing, Tai Chi, jogging, jumping, running, dancing, vibration platform, strength training and a combination of the exercise above. There was a high variability in the frequency of the exercise interventions ranging from three to six times a week. This makes it evident that there is no agreement in a standard way of testing physical activity for prevention and treatment of osteoporosis, which makes the interpretation of multiple studies difficult. Yet there is a small statistically significant effect of exercise on BMD, which is important evidence in clinical practice towards promoting research in ways to determine optimal exercise programme (Howe et al., 2011). According to another review by Cheung and Giangregorio (2012), literature shows that the most adequate interventions involve progressive resistance training with walking or aerobic dancing if willing to improve spinal BMD.

Daniels (2008) has identified physical activity that improves BMD: aerobics, strength training, yoga, pilates and feldenkrais. However, Daniels (2008) points out that exercise should be performed under the recommendation of a physician while maintaining the natural curvature of the spine and a good neck alignment throughout any physical activity. Daniels (2008) also recommends wearing a backpack or weighted vest during walking or hiking in order to improve the effects on bone. If due to spinal osteoporosis wearing a weighted vest is not possible, Daniels (2008) recommends wearing a weighted belt. However, it is concerning that no specific evidence is

given for the effectiveness of wearing a weighted belt. According to Van Norman (2010) older people with osteoporosis should avoid excessive flexion of the spine, ballistic or jarring movements and standing for too long in only one leg with severe osteoporosis. Chair exercise may also be more appropriate than standing. Exercise that Van Norman (2010) recommends are moderate weight bearing, low impact aerobics and vigorous walking given a physician available to support any exercise class for safety.

Research looking at physical activity aimed at those with osteopenia is sparse. For instance, Eriksen (2012) only mentioned 'regular exercise' as an alternative intervention to osteopenia. It is clear that general statements as this are not aimed to help professionals and the general population towards the prevention and treatment of osteopenia. Thus for the treatment of osteopenia it is not clear whether to follow the same recommendations as for older healthy adults or for the osteoporotic subjects.

Vainionpää et al. (2006) showed that accelerometers could be employed to measure the intensity of physical activity and its relationship with lumbar BMD determined through DXA. Vainionpää et al. (2006) suggested that high accelerations of 5.4 g (52.9 m/s^2) had positive effects on the lumbar spine of healthy premenopausal women yet more research is needed to validate this technique for the older population with osteoporosis. Vainionpää et al. (2006) tested high impact activities that may be contraindicated for osteoporotic subjects such as running, jumping and drop jumping. This study suggested the use of accelerometry to objectively measure intensity of physical activity. However, it only measured peak acceleration and BMD, which do not provide a full picture of bone structure.

It has also been suggested that exercise mediated alterations on the production of less serum sclerostin may facilitate the maintenance of bone mass (Amrein et al., 2012). Further research is needed to determine if the effect of exercise on the production of this protein will further support the prescription of exercise for the prevention and treatment of osteopenia and osteoporosis. Similarly, Maïmoun and Sultan (2013) have suggested that biochemical markers of bone turnover

could clarify the effect of exercise on bone metabolism, providing a possible way to standardize an optimal exercise program for all ages and bone health. Yet further research is required before transferring this method to the clinic. In addition, current research into bone turnover markers fail to address interest on vertebral fractures (Biver et al., 2012). An additional disadvantage of biochemical markers is that these are not specific to skeletal sites.

2.2.4.3. Synopsis of physical activity and exercise

Physical activities that have been identified to either produce a modest improvement for the bone quality in the spine or just preserve it are sparse (walking, volleyball, Tai Chi, aerobics, strength training and a combination of physical activities), this is due to the inability of DXA to take into account bone structural changes (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Gremeaux et al., 2012, Cheung and Giangregorio, 2012). In addition, there is a lack of research based guidelines for how often and how much exercise is optimal for osteoporotic subjects to respond safely and positively to exercise (Kohrt et al., 2004, Hamilton et al., 2010a, Bergmann et al., 2011, Kasturi and Adler, 2011, Gremeaux et al., 2012, Cheung and Giangregorio, 2012, Mayer et al., 2011). This might be the result of conflicting definition of physical activities combined with the lack of a way to measure physical activity objectively. Some activities are classified according to measured expenditure of calories, others by intensity and often by their aerobic characteristics. Classifying physical activity based on these measurements leads to overlap, therefore it is necessary to find a practical technique to measure the effect of physical activities on bone noninvasively. This practical technique must be able to measure the effect of physical activity not only in relation to bone density but also in relation to size and structure.

2.2.5. Interaction between mechanical and non-mechanical stimuli

Srinivasan et al. (2011) warned that finding optimal exercise regimens that sufficiently maintain and increase bone strength on the older osteoporotic population may not be possible to achieve unless age related bone cell function and signalling deficits are targeted. There is no clear evidence of how bone will respond to physical activity while being subjected to pharmacological treatments influencing osteoclasts and osteoblasts activity. For instance, the combination of alendronate and

WBV (12.6 Hz, 3 cm, knees flexed 60°, 6 bouts of 1 min for 8 months) does not change lumbar spine BMD (Gusi et al., 2006). A new clinical trial is being performed in order to test individually designed treatments for osteoporosis combining pharmacological and non pharmacological treatments (Edmonds et al., 2012). It is expected that this will give evidence on the impact of a more complex and complete approach to improve bone health. In general, before specific skeletal sites can be targeted, supplementation of exercise will require the optimisation of preclinical models and clinical trials.

2.2.6. Summary

A large proportion of people do not receive preventive treatment for osteoporosis. In addition, the type of osteoporosis treatment varies across geographic regions (Northern Europe, Australia and USA) showing inconsistency and promoting lack of effectiveness of pharmacological treatments (Díez-Pérez et al., 2011). Pharmacological and surgical interventions are generally used when osteoporosis is severe and vertebral fractures are present. Routine screening and diagnosing may allow for non-pharmacological interventions if detected at an early state. However, the long term benefits of treating osteopenia through diet, physical activity or whole body vibration are not known. In addition, the efficacy and safety of whole body vibration has yet to be shown. Some physical activities have been identified to either produce a modest improvement of spinal BMD or just preserve it. It is necessary to find a way to objectively measure physical activity regardless of geographical region, BMD and age. This will enable future research to determine the optimal exercise prescription that will ideally include frequency, intensity, type, time and progression according to individual capabilities and current health status while addressing expectations and barriers.

2.3. Response of the spine to vibration

Vibration can be delivered to the human body from different sources. The effects of vibration on the body have been studied mainly for safety purposes in occupational settings and for seated persons. Studies on the treatment of sarcopenia and osteoporosis have provided information on the effects of vibration mostly in animal studies and through whole body vibration. In this section,

physiological effects of vibration on the human spine will be presented along the importance of the spinal curvatures and safety aspects.

2.3.1. Physiological effects

Nearly all cell types (myocytes, platelets, endothelial cells, chondrocytes, fibroblasts and bone cells) are able to sense and respond to physical factors in their environment (Thompson et al., 2012). Mechanotransduction is the process by which cells convert environmental signals (mechanical stimulation) into biochemical signals (Thompson et al., 2012). At least four types of cells are believed to be mechanosensitive in bone: osteoclasts, osteoblasts, osteocytes and mesenchymal stem cells (MSC) (Thompson et al., 2012, Chen et al., 2010). The theory states that vibration may amplify mechanotransduction signals by altering intramedular pressure (Qin et al., 1998), promoting fluid flow (Qin et al., 2002, Chen et al., 2010, Letechipia et al., 2010), causing a cyclic contraction and relaxation of muscles (Judex and Rubin, 2010), increasing growth hormone and testosterone concentrations (Totosy de Zepetnek et al., 2009) and by inducing progenitor cells to become bone cells instead of fat cells (Ozcivici et al., 2010).

Through this intricate capacity of bone cells to respond directly and indirectly to mechanical and biochemical signals, the skeleton is capable of adapting (change its morphology) to new functional demands (Chen et al., 2010).

2.3.2. In vitro studies

The mechanotransduction of bone cells is studied in vitro in order to have a greater control over mechanical signals. However, in vitro study of bone cells is subjected to a paradox in which larger strains (compared to in vivo studies) and low frequencies (0.1 to 1 Hz) induce physiological changes (Thompson et al., 2012). This may be explained by the multiple disadvantages of in vitro studies. For example, signal pathways present in vivo may not be replicable in vitro. Cell culture is far from real physiological conditions due to the two dimensional setting and limited time for any experiment (due to cell survival outside normal conditions). The ability of cells to sense mechanical signals depends on mechanosensors, which can be a molecule, a protein complex or a

biological structure located for example in the extracellular space or on the plasma membrane (Chen et al., 2010, Thompson et al., 2012). Thompson et al. (2012) concluded that it still remains unclear how mechanical signals are sensed by cells. There are many theories of possible mechanoreceptors that are not the topic of this thesis. Instead, the focus will be on the measurement of the functional musculoskeletal environment of the spine in vivo.

2.3.3. Animal studies

The natural mechanical environment in which bone cells survive consists of multiple physical factors such as stress, strain, pressure, fluid flow, streaming potentials and acceleration. Components of these factors are also present and measurable: magnitude, frequency and strain rate (Thompson et al., 2012).

Numerous animal models (dogs, rats, mice, sheep, turkeys, roosters, horses, geese and rabbits) have provided evidence that bone tissue can be loaded in a controlled manner (controlled vibration) in order to form new bone (Rubin et al., 2001, Skerry, 2008, Chen et al., 2010, Thompson et al., 2012). This vibration is controlled in terms of type, intensity and duration as well as its effect in terms of bone gain or loss. Load intensity can be measured in terms of strain magnitude as well as strain rate (Kohrt et al., 2004). Duration refers to how often a specific type of stimulation (with a specific strain magnitude and strain rate) must be provided and frequency refers to the amount of rest between each stimulation. Animal studies have provided information on the frequency and strain experienced during daily activities. Low frequency (1 to 3 Hz), high frequency (10 to 50 Hz from muscular contraction) and low magnitude (< 5 με) have been observed (Thompson et al., 2012). Interestingly, studies have also suggested that some mice are genetically more sensitive to vibration and others genetically nonresponsive (Prisby et al., 2008) though this has not been tested in humans. Other animal studies have provided evidence that there is a threshold at which the bone is not responsive to stimulation (Kohrt et al., 2004). For stimulation, animal studies have used low vibration magnitudes (from to 0.25 g to 3 g) and high frequency (20 to 90 Hz) (Totosy de Zepetnek et al., 2009). These studies have suggested that muscle and bone do not respond to vibration in the same frequency and that bone response is not dependant on vibration amplitude (Judex and Rubin,

2010). More controversial is the suggestion that the skeleton may respond to vibration regardless of strain and muscular contraction. Animal studies have given evidence that bone responds to motion of very small magnitude, equivalent to a small amount of acceleration (Garman et al., 2007). Given the ability of bone to respond to vibration in the absence of muscular contractions and deformation, it may be possible that the high percentage of transmissibility from the appendicular skeleton to the axial skeleton is highly sensed by bone cells (Judex and Rubin, 2010). However, there is insufficient evidence to translate this for human application (Rubin et al., 2001, Kohrt et al., 2004, Skerry, 2008). In general, though it is difficult to translate animal studies to human application, it has been recognized that mechanical stimulation has a potential for its use as a treatment for sarcopenia and osteoporosis.

2.3.4. Human studies

The effects of vibration on the body have been studied in different ways depending on the origin of the vibration. For instance, vibration can be artificially produced by a motor or naturally produced during gait. In addition, information on the effect of artificial vibration on the body is mainly oriented to the upper limbs, lower limbs and the whole body but very seldom to the spine.

Human gait produces vibration that propagate through the human skeleton and soft tissue (Voloshin et al., 1981, Wosk and Voloshin, 1981, Voloshin and Wosk, 1982, Smeathers, 1989a, Smeathers, 1989b, Kim et al., 1993). However it is not clear if this mechanical input directly stimulates bone formation paths or if muscular contractions, in response to this input, signal bone formation (Robling, 2009) or if vibration only increases muscular mass which later exposes bone to greater forces achieving bone formation (Judex and Rubin, 2010). Nonetheless there is strong evidence that ground reaction forces during gait are inherently linked to bone maintenance and formation due to the severe consequences observed when removing gravitational force (Judex and Carlson, 2009). The mechanism of action of vibration stimulation of bone during physical activity may have multiple pathways.

Artificially produced vibration through mechanical actuators and delivered to the human body, through a vibrating platform, as part of an exercise training routine is another mode of exposure. Though the physiological effects are not yet well understood (Griffin, 1990, Rittweger, 2010). This mechanical oscillation will affect muscle, tendons, nerves, skin and bone with responses in terms of temperature, metabolic changes, perfusion alterations and hormonal changes (Griffin, 1990, Rittweger, 2010). Through FEA it has been suggested that every tissue of the spine responds differently to vibration with consequences on fluid perfusion being dependant of vibration frequency, amplitude and duration (Prisby et al., 2008). Muscles are exposed to cyclic transition between concentric and eccentric contractions. A reflex contraction in muscle, which is not well understood, also occurs. Due to the enhanced muscular contractions, energy demand is also increased as well as intramuscular temperature (due to the transformation of mechanical energy to into heat). There is an increment in muscle and skin blood flow due to the increased need of energy for the enhanced muscular contraction. There are also controversial suggestions that vibration improves joint stability as well as flexibility and balance. Hormonal responses to vibration exercise have also been documented for testosterone, growth hormone (IGF axis), cortisone and catecholamine (Prisby et al., 2008, Rittweger, 2010). Finally, there is no sufficient research evidence to conclude in the effects of vibration on blood lipids and glucose (Rittweger, 2010).

Studies comparing the effects of vibration exercise with normal exercise have been performed mostly in athletes and healthy adults (Rittweger, 2010). Yet, there is insufficient evidence to determine if one is better than the other. Another use of vibrating force plates is for clinical applications, for example for improving muscular frailty, for central nervous disorders and for the treatment of osteopenia and osteoporosis.

It is possible that vibration stimulation is moderately effective in increasing BMD and bone strength by a combination of all the bone responses described through animal and in vitro studies. Judex and Rubin (2010) pointed out that in order to understand the mechanism through which vibration stimulates bone, it is necessary to determine the characteristics of the ‘mechanical

environment' generated by that vibration stimulation at different 'levels and hierarchies' (for example environment and hierarchies may be dissimilar between lower limbs, upper limbs and spine) (Kohrt et al., 2004).

The effect of mechanical stimulation has been mainly evaluated in large bones (tibia, femur, radius, ulna) and hand, rather than on the spine because it is a location difficult to measure and stimulate (Griffin, 1990, Mansfield, 2005a). Human studies have mainly focused on vibration produced on the working environment (Griffin, 1990, Mansfield, 2005a), vibration as treatment for bone loss due to prolonged rest (Garman et al., 2007), spinal cord injury (Asselin et al., 2011), space flight (Judex and Carlson, 2009, Clément et al., 2010) and fewer for the treatment of osteopenia and osteoporosis (Rubin et al., 2001, Rubin et al., 2003, Rubin et al., 2004, Wysocki et al., 2011). The efficacy and safety of artificially delivered whole body vibration for the treatment and prevention of osteoporosis will be outlined in detail later. In general, the interaction of frequency and intensity of loading has not been fully understood (Kohrt et al., 2004).

Regarding the effect of vibration on the intervertebral disc, studies on the effect of artificially delivered vibration (for exercise purposes and during work through vibrating machinery) have found that the height of intervertebral discs change in a non-uniform way and experience fluid flow volume decrement (Hill et al., 2009). The vertebrae are believed to 'move' stretching the soft tissue attached to them (ligaments, tendons and muscles) (Hill et al., 2009). Kiiski et al. (2008) delivered sinusoidal vibration though a platform to standing subjects in order to explore the transmissibility of the human body. It was found that transmission of vibration may be modified by frequency and magnitude but acknowledged that it is a complicated phenomenon due to the nonlinear nature of the musculoskeletal system. The key contribution of this study is that it attempted to translate animal and in vitro studies to human application while performing measurements at the spine as well. Kiiski et al. (2008) found that the lumbar spine peak acceleration was amplified especially after 10 Hz and that transmission of vibration should be studied specifically in the elderly because it can vary due to sarcopenia and stiffer tendons and muscles. Most importantly, it was mentioned

that the safe delivery of vibration, such as that seen in animal studies, for the frail human must be assessed. Nonlinear characteristics of transmissibility of the spine were observed also by Mansfield and Griffin (2006) when exposing seated subjects to whole body vibration. It was hypothesized that this nonlinear response was due to the muscular response to vibration but more likely by a combination of factors.

Overall, more research is needed to determine if WBV has any positive long or short term effect on the human spine before its use in the clinic. Little is known of the effects of vibration produced during gait on the healthy spine and nothing on the osteoporotic one. For these reasons, artificial vibration is not fully recognized as part of a treatment for osteoporosis in contrast with vibration produced during physical activity. Current evidence on the effect of physical activity for the treatment of osteoporosis was presented in section 2.2.4.

2.3.5. Spinal curvature

There is currently no information about how lumbar and thoracic spinal curvatures will affect vibration transmissibility during gait or during daily life physical activities. When the spine of a healthy adult is subjected to WBV it has been suggested that the flattening of the lumbar lordosis increases the transmissibility of forces (Bazrgari et al., 2008). In another study, it was suggested that spinal curvature changes, due to older age, lead to a decrement of the fibre angles of lumbar extensor muscles in older adults when compared with young adults. This may affect compressive and shear loads in the lumbar spine (Singh et al., 2011). Therefore it would be reasonable to think that vibration transmissibility is significantly affected during gait given a significant change on spinal curvatures and given changes in the angle of lumbar extensor muscles due to older age. In contrast, the clinical view is that the spinal curves increase the strength of the overall spine and help the body balance and absorb shocks during gait (Tortora and Grabowski, 2003). However, this hypothesis needs to be tested for the older healthy and older osteoporotic population in the present study.

2.3.6. Safety aspects

The International Organization for Standardization (ISO) has issued the standards on the limits of whole body vibration (ISO, 1997) and hand vibration (ISO, 2001) in industrial settings. No standard exists on the limits of vibration exposure employed as means of exercise and therapy. Rubin (2006) warned that vibration of magnitude above 9.81 m/s^2 is not considered safe for human exposure. It is concerning that companies that manufacture exercise training vibrating force plates as well as vibrating force plates for the treatment of osteoporosis, have produced devices that greatly exceed this threshold. In a similar way, the standards issued by the ISO have also warned of the risks of exposing the human body to frequencies in the 20 to 50 Hz range. Rubin (2006) stressed that other frequencies and amplitudes of vibration (not considered safe) can damage vertebral discs, cartilage, ligaments, tendons, nerves and the cardiovascular system. Since there is no other study exposing commercial force plates as Rubin has done, these warnings may be biased. Vibration plates should be used with caution given that there is no organization providing a standard for their use.

International standards on the use of vibration for exercise and clinical applications will emerge until more research evidence is available on the safety, effectiveness and long term effects of vibration for these applications.

2.3.7. Summary

Most of the studies that explain structural bone changes due to mechanical stimulation are in animal models, thus there is the need to translate these findings for human application through clinical trials. Mechanical stimulation is capable of enhancing the response of bone cells through biochemical paths that are not yet well understood. Mechanical signals, such as vibration produced during human gait, offer a non invasive and non pharmacological alternative for the treatment of osteopenia and osteoporosis, though the development of a clinical intervention requires the characterization of the bone's mechanical environment.

2.4. Methods to assess spinal osteoporosis

In this section the common clinical techniques used to detect and monitor osteoporosis in vivo will be presented along with new techniques that are still under research, development and in clinical trials. Focus will be given to vibration transmissibility which is a technique that has potential to monitor the effects of osteoporosis in vivo (during daily life physical activities).

2.4.1. Methods currently used in the clinic

X-Ray absorptiometry assess the amount of mineral in an area of bone. It is BMD in the area rather than volumetric measure. Single X-ray absorptiometry (SXA) is done at the heel and forearm. Dual X-ray absorptiometry (DXA) also known as dual photon absorptiometry (DPA) is done at the spine and hip. DXA is regarded as the "gold standard" since this technique has been thoroughly validated and widely used in routine clinical settings (WHO, 2003). The distribution of bone mineral content of a young and healthy adult population has a normal distribution. Individual BMD values are expressed in terms of standard deviation units in relation to that normally distributed population, this value is called T-score. There are four general diagnostic categories based on the T-score: normal (T-score $\geq$ -1), osteopenia (-2.5 < T-score < -1), osteoporosis (T-score $\leq$ -2.5) and severe osteoporosis (T-score $\leq$ -2.5 plus a fragility fracture).

The largest source of error for the DXA result is soft tissue density and heterogeneity of that tissue due to previous fractures or osteoarthritis, these factors can erroneously increase the BMD measurement (WHO, 2003, Griffith and Genant, 2012). Increased body and marrow fat can erroneously decrease the BMD on the spine and hip (Griffith and Genant, 2012). Recent techniques developed, that will be outlined in the next section, have found that BMD do not quantify for factors that contribute to bone strength (tissue properties, morphology, microarchitecture) and that other methods are better for fracture prediction as well as for monitoring and assessing the response to treatment (Bouxsein, 2011).

Broadband ultrasound attenuation (BUA) and speed of sound (SOS) at the calcaneus (heel) bone provide an approximation of the T-score that would be obtained from a DXA scan. It has the

advantage of involving non-ionizing measurements and being a low cost equipment (NOF, 2010). Normal bone has a higher attenuation than osteoporotic bone, in other words, the more complex the structure of the bone, the more sound will be blocked (WHO, 2003). Although QUS does not measure BMD directly, BUA can be used clinically to predict vertebral fractures risk based on the approximation of the T-score. BUA has been shown to be of predictive value in the assessment of osteoporosis by both retrospective and prospective studies (Bauer et al., 1995, Bauer et al., 1997).

2.4.2. Methods under research and development

Computed Tomography (CT) uses X-ray technology to obtain cross sectional images of a body. It can visualize bone with high spatial resolution. A CT scan is a mathematical representation of a body rather than an image. Quantitative CT (QCT) measures true volumetric density in comparison with DXA. This technique is more suitable for monitoring treatment since trabecular bone is more responsive. Multi detector CT (MDCT) can provide texture of trabecular bone of the spine and femur. Drug treatment effects are also better monitored with microarchitecture of spinal bone provided by MDCT (Bauer and Link, 2009, Griffith and Genant, 2012). Another major contribution of MDCT is that volumetric BMD (vBMD) can be determined. For example, Samelson et al. (2012) used QCT to determine vBMD, geometry and strength of thoracic and lumbar sections of 690 human volunteers. Samelson et al. (2012) concluded that age related changes were greater in the lumbar spine than in the thoracic region (also greater in women than in men). Studies considering the whole spine as a homogenous section would be failing to explain the true influence of degeneration of the spine on its biomechanics.

Micro CT (μCT) was the first system to provide the "true" 3D trabecular architecture increasing our knowledge on trabecular network. As a result of this system, bone loss was seen for the first time transforming trabeculae form plate-like to rod like structure, rather than thinning of trabeculae as previously thought (Stauber et al., 2006). An additional structural characteristic that CT offers is the measurement of cortical porosity, however this technique has also been tested only in the femur and not vertebrae (Bousson et al., 2000). Since it is necessary to have a CT unit at lower cost and lower radiation dose for clinical practice, the peripheral (p) skeleton was chosen to be studied and

to mirror parameters of vertebrae and femur through high resolution pQCT (HR-pQCT). Disadvantages of HR-pQCT systems are that it is only available for distal radius and distal tibia and measurements are sensitive to movement artifact (Griffith and Genant, 2012).

The major disadvantages of QCT are the high radiation exposure, difficulties with quality control and high cost of the equipment (WHO, 2003, Bauer and Link, 2009).

Magnetic Resonance Imaging (MRI) is a non-ionizing imaging method and has the added ability of determining physiological features of bone such as marrow fat content, marrow diffusion, marrow perfusion and water content (Griffith and Genant, 2012). In short, bone structure and metabolism can be determined simultaneously (Bauer and Link, 2009). Similar to CT, MRI can help determine cortical porosity but based on water content, though on tibia not vertebrae (Techawiboonwong et al., 2008). Drawbacks of MRI are that it provides no direct information on density but resolution of the internal structure of trabecular bone. This technology is still costly, is limited by signal to noise ratio, has resolution issues, the measurements not only require considerable time to be performed but their analysis is complex and demanding (Griffith and Genant, 2012). Most in vivo MRI studies have focused their attention on distal radius and tibia as well as the calcaneus for the study of osteoporosis. High resolution MRI (HRMRI) has been employed to determine structural parameters of distal radius which are better than DXA distinguishing women with and without vertebral fractures, but still would be desirable to see this technique applied directly to the spine (Krug et al., 2008). An interesting MRI study has shown how perfusion is reduced in osteoporotic vertebrae in comparison with those with normal BMD (Griffith et al., 2006). But it would be more interesting and useful to see in the future this technique detecting perfusion changes before vertebrae become osteoporotic or osteopenic.

So far, CT and MRI measurements while subjects are in a static position are performed. Before this technology can be applied to study the dynamics of bone microstructure in vivo, it requires to be portative. pQCT is an example of how this technology is being developed towards portative

systems but will require further research before being used in clinical practice for the study of the spine.

When combining areal BMD (DXA) with volumetric X-ray absorptiometry (VXA), the prediction of failure given a determined force is significantly improved in comparison with employing BMD alone and it also correlates well with vQCT (Ahmad et al., 2010). However, this technique has only been tested for the femur and not spine.

Combining vQCT and Finite Element Analysis (FEA) has provided information on how vertebral strength decreases with age twice as much in women than men. Results show that this is due to the fact that women have a greater decline of cortical bone strength while trabecular bone decline is similar in both sexes (Christiansen et al., 2011).

Combining MRI, WASPI pulse sequences and phosphorus spectroscopy (^{31}PMRS) shows that it is possible to measure in-vitro bone matrix mineralization, a technique that could help to differentiate between osteoporosis and osteomalacia (Wu et al., 2010). On other study, Pothuaud et al. (2002) combined MRI and FEA to increase the prediction of elastic moduli of trabecular bone from third lumbar vertebrae. The disadvantage of this study is that it was an in-vitro study, thus it is not available for clinical use.

More recently, a series of studies have employed FEA using geometric and material properties previously obtained through in-vitro tests, CT and MRI images to study the effect of regular daily activities on biomechanics by creating a whole model of the lumbar spine (Schmidt et al., 2006, Schmidt et al., 2007, Schmidt et al., 2010). The fact that this study looked at physical activity rather than single independent mechanical measurements is what makes it different and useful for understanding the behaviour of the spine in real life while employing previous research. This model was able to demonstrate how fluid pressurization and flow direction within the spine have an important role in determining the biomechanical response during loading and recovery phase (such as during daily life physical activities) (Schmidt et al., 2010). Even though this model is helpful, it

is not answering the question of how osteoporosis alters this biomechanical response during daily life physical activities.

Vibration testing has been employed in combination with other techniques. Cheung et al. (2003) created a FEA model from a L4-L5 segment previously scanned by CT. Cheung et al. (2003) included in the model material properties from previous literature and added a poroelastic model for taking into account permeability of fluid. Vibration loading was sinusoidal with magnitudes from 180 to 420 N (for a 70kg person the static load is 1000 N) at frequencies from 0.5 to 4 Hz. The model gave evidence that vibration increases the disc fluid exchange however it does not give an insight into how prolonged vibration (such as during daily life physical activities) would affect the biomechanical response of the spine.

Regarding the reliability of FEA, these models often consider a single fixed geometry of either a single subject's spine or a single element of it. This means that these models cannot account for effects due to natural variability between subjects. Niemeyer et al. (2012) recommended a minimum sample size of 100 in order to accurately predict the lumbar spine's response to loads through FEA. Some geometrical variables that were identified to be truly important for FEA of the lumbar spine were related to disc geometry and facet's position (Niemeyer et al., 2012). Thus the biomechanical results derived from FEA should be used with caution.

Mc Donnell et al. (2009) μCT scanned a human lumbar vertebrae and created a 3D model in stereolithography (3D printing technology). Through an algorithm, Mc Donnell et al. (2009) simulated bone loss and performed dynamic compression tests in a wide frequency range. Mc Donnell et al. (2009) found a relationship between bone loss, resonant frequency, apparent stiffness and strength. However, this technique is only valid for in-vitro tests and it requires the knowledge of the input stimulus, which is unknown during daily life physical activities.

2.4.3. Vibration transmissibility

Vibration refers to the oscillatory motion of a body or system. Any system having a mass and any degree of elasticity is capable of vibration (Thomson, 1993). The human body viewed as a system

is subjected to oscillatory motion during gait. This oscillatory motion causes stress waves that propagate through the human skeleton and soft tissue (Voloshin et al., 1981, Wosk and Voloshin, 1981, Voloshin and Wosk, 1982, Smeathers, 1989a, Smeathers, 1989b, Kim et al., 1993). The nature of those stress waves or vibration can be considered as free, forced, damped, random and not stationary (Griffin, 1990, Thomson, 1993, Mansfield, 2005a). During gait the human body may be under free vibration since there are forces generated within the system. It could also be considered to be under forced vibration since every heel strike (or shock) is an external force that maintains vibration as long as the body is moving. It is also damped vibration since the human body removes energy by friction (between articulations and soft tissue) and by viscous drag of fluid (inside cartilage tissue, bone tissue and intervertebral discs).

The human body is inherently highly damped (Mansfield, 2005a). For instance, previous studies have confirmed that the lower limbs attenuate transient impulsive forces during gait. Chu et al. (1986) observed that cadaveric lower limbs attenuated 59% of 19 g impulsive forces at 0.5 Hz. Chu et al. (1986) also noticed that pathological changes in the knee reduced the attenuation capacity, potentially exposing the hip and spine to higher impulsive loads. Through frequency analysis, Angeloni et al. (1994) found that damping of heel strikes increases from feet to head. Human gait may produce random vibration because successive heel strikes are different (hence the possibility of nonlinear characteristics) and it is not constant (or stationary). Previous research has found that human gait produces acceleration with several magnitudes and frequency components and that these are different according to location of the body (Rao and Jones, 1975, Antonsson and Mann, 1985, Cappozzo, 1982, Angeloni et al., 1994). Rao and Jones (1975) employed accelerometers to determine that walking produces vibration with the greatest frequency contents between 1.2 and 2 Hz. Similarly, Rao and Jones (1975) found that acceleration and deceleration forces decrease significantly after 50 Hz. Antonsson and Mann (1985) employed a force platform to measure the spectral content of human gait and determined that 98% of the power was contained below 10 Hz. Cappozzo (1982) employed a photogrammetric technique to measure displacement at the trunk and calculated acceleration. From spectral analysis Cappozzo (1982) detected four harmonics between

0.75 and 4.8 Hz. Angeloni et al. (1994) analysed the frequency content of gait through cameras and found a maximum of 9.8 ± 1.5 Hz at the shank, 9.2 ± 1.5 Hz at the trunk and 7 ± 1 Hz at the head. Not only human gait produces complex vibration with different amplitudes, frequencies and phases but the response of an individual exposed to this complex vibration will depend on the magnitude and frequency of the stimulus due to the inherent heterogeneous mechanical properties of the human body.

Any vibration is measurable through displacement, velocity and acceleration (Thomson, 1993). Acceleration can be measured through inertial sensors. Accelerometers are employed to measure periodic acceleration and deceleration of a body, namely vibration (Griffin, 1990, Mansfield, 2005a). Gyroscopes are employed to measure tilt and rotation. The study of vibration transmission through the human body has been approached through impedance, transmissibility and modelling methods. Most common impedance methods are transfer impedance, apparent mass and absorbed power (Griffin, 1990, Mansfield, 2005b). Transfer impedance ($z(f)$) consists in the ratio of the force ($F(f)$) to the velocity ($v(f)$) in the frequency spectrum (f) (Equation 2.4-1). Velocity is calculated from acceleration and $z(f)$ has to be normalized by a single mechanical impedance value at the lowest frequency with the highest coherence. Apparent mass ($M(f)$) consists in the ratio of the force to the acceleration ($a(f)$) in the frequency spectrum (Equation 2.4-2). Apparent mass requires normalization by the weight of the subjects in the static seating or standing position. Absorbed power ($P_{abs}(f)$) consists in the product of the modulus ($\left|G_{fv}(f)\right|$) and phase ($\cos\phi_{fv}(f)$) of the cross spectrum between force and velocity in the frequency spectrum (Equation 2.4-3). For compliant systems (human body) absorbed power is a function of vibration magnitude and mass. Absorbed power also requires normalization by the weight of the subjects in the static seating or standing position. In general, impedance methods provide data on frequencies of vibration to which the human body is most mechanically sensitive. Impedance measurements require the knowledge of a driving force and acceleration, which would be measured at the surface supporting the body in a static position. Hence impedance methods are widely used when studying

the spine exposed to vibration for seated and standing subjects (Griffin, 1990, Mansfield, 2005a, Mansfield, 2005b).

$$z(f) = \frac{F(f)}{v(f)} \quad \textbf{2.4-1}$$

$$M(f) = \frac{F(f)}{a(f)} \quad \textbf{2.4-2}$$

$$P_{abs}(f) = \left|G_{fv}(f)\right| \cos \phi_{fv}\,(f) \quad \textbf{2.4-3}$$

Transmissibility is the ratio of vibration measured between two different points in the frequency spectrum, also called transfer function. The transfer function consists in dividing the power spectral density (PSD) of the output vibration to the PSD of the input vibration (Equation 2.4-4). The PSD method is useful for both linear and nonlinear systems. Another method involves calculation of cross-spectral density (CSD) but assumes that the system is linear (Equation 2.4-5). Transmissibility can be presented in a plot where the x-axis is frequency (Hz) and the y-axis is transmissibility (which has no units). An example of the response of a simple dynamic system to vibration can be seen in Figure 2.3-1. Transmissibility values above unity mean amplification while below unity mean attenuation. Transmissibility of magnitude one means that 100% of vibration is transmitted (Figure 2.3-1). With greater damping in the system, the peak transmissibility response decreases (Mansfield, 2005a). At the resonance frequency the response is greater than the stimulus (Mansfield, 2005a). Thus transmissibility at specific frequencies gives information on stored energy or dissipated energy by bone and soft tissue between the two points measured (Griffin, 1990, Mansfield, 2005a, Mansfield, 2005b).

$$T(f) = \sqrt{\frac{PSD_{output}(f)}{PSD_{input}(f)}} \quad \textbf{2.4-4}$$

$$T(f) = \frac{CSD_{input.output}(f)}{PSD_{input}(f)} \quad \textbf{2.4-5}$$

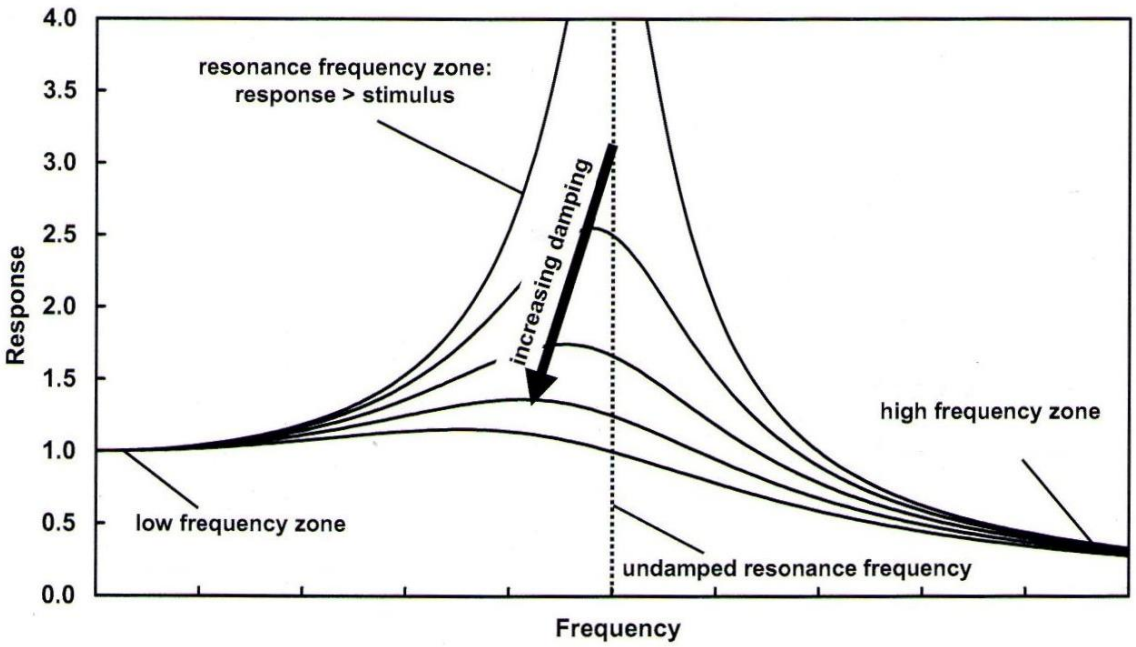

Figure 2.4-1 Response of a simple dynamic system to vibration. Modified from (Mansfield, 2005a)

Since vibration transmissibility does not require knowledge of a driving force supporting the body during a static position, it is an ideal technique to measure transmissibility of the human spine during physical activities by attaching accelerometers over spinous processes. Although impedance methods provide desirable information, these methods cannot be employed for the human body during physical activity. Furthermore, an additional advantage of measuring vibration transmissibility is that it is not affected by gender nor anthropometric measures (Griffin and Whitham, 1978, Mansfield, 2005a). For instance, Griffin and Whitham (1978) measured seat to head vibration transmissibility (from 4 to 16 Hz) in men, women and children and found no significant difference between the three populations. Similarly, Mansfield (2005a) has indicated that there is no evidence of predicting differences in transmissibility between subjects from variables such as anthropometric measures, while it is know that anthropometric measures are correlated to gender.

The magnitude of vibration as acceleration can provide additional information in terms of root mean square acceleration and frequency content. If a vibration is measured for an infinite duration

or for a reasonable number of cycles at all frequencies, root mean square acceleration (RMSa) will represent the magnitude of the vibration. When measuring acceleration transmitted to different locations over the human body the effect of gravity is removed, then RMSa is equivalent to its standard deviation (Equation 2.4-6). In addition to RMSa, spectral analysis of acceleration gives information on the energy present at all frequencies, thus transmissibility through the PSD method is commonly presented parallel to PSD of acceleration signals (Griffin, 1990, Mansfield, 2005a).

$$RMSa = \left[\frac{1}{N}\sum a^2(i)\right]^{1/2} \qquad \textbf{2.4-6}$$

Vibration transmission through the human body in-vivo has been studied mainly through the employment of accelerometers. Other studies have employed alternative technology to study stress wave propagation such as strain gauges in-vitro, electromagnetic fields and 3D laser vibrometers in-vitro. Pelker and Saha (1983) glued strain gauges to human cadaveric long bones and measured the propagation of a stress wave. Pelker and Saha (1983) found a parabolic relationship between the porosity of bone and its attenuation coefficient. The main disadvantages of employing strain gauges are their invasive nature and that the effect of surrounding soft tissue on the measurements is not known.

Due to the piezoelectric nature of bone, electromagnetic fields may be employed to study stress wave propagation. Ahmed and Abd-Alla (2002) showed through a numerical model that the magnetic field created by the electrical charge motion generated by a long bone under a stress wave, can be represented by a plot of variation of that magnetic field as a function of cylindrical coordinates representing the dimensions of the bone. The work of Ahmed and Abd-Alla (2002) is promising in terms of the non-invasive nature of the method based on electromotive forces but is a theory still in development with no current device available for in-vivo trials. Laser vibrometry is able to measure small vibration without the need of contacting the surface of bone. Rixen and Schuurman (2012) were able to measure vibration produced by heart pulses on the surface of the

thorax and abdomen in-vivo. However, there is no technique available to approximate vibration under the skin through laser vibrometry in vivo.

Studies on spinal vibration transmission through accelerometers have focused on vibration hazards during work for seated persons in relation to low back pain (Panjabi et al., 1986, Pope et al., 1987, Collins and Whittle, 1989), osteoarthritis (Collins and Whittle, 1989) or in order to provide data for biomechanical models (Matsumoto and Griffin, 1998, Fritz, 2000, Pankoke et al., 2001, M-Pranesh et al., 2010). These studies have employed transmissibility and apparent mass methods to study the effects of sinusoidal vibration and random vibration, produced by vibrating seats, on the spine. Panjabi et al. (1986) surgically inserted stainless steel pins (Kirschner k-wires) into the spinous processes of two lumbar vertebrae and the sacrum. Application of Kirschner wires for clinical use in order to study daily life physical activities is not feasible due to their invasive nature. As an alternative, it is possible to employ a method to correct for skin movement when attaching sensors to the skin either with glue or adhesive tape (Kitazaki and Griffin, 1995, Matsumoto and Griffin, 1998, Pankoke et al., 2001). This skin correction method will be outlined later.

Other studies based on stress wave propagation have employed techniques of structural health monitoring (used in civil engineering) to determine the mechanical properties of human bone in-vitro (Keller and Colloca, 2007, Kawchuk et al., 2009, Bediz et al., 2010, van Engelen et al., 2011, van Engelen et al., 2012), in-vivo (Keller et al., 2000, Bediz et al., 2010, Bhattacharya et al., 2010), to monitor fracture healing (Akkus et al., 1998) and also to provide data for biomechanical models (Nakai et al., 2007). These studies have also benefited from the use of non-invasive techniques such as attaching accelerometers to the skin with adhesive tape and employing a skin-sensor interface movement correction method. Despite the fact that structural health monitoring provides the opportunity to perform non-destructive sensing and obtain data on presence, location, severity and prediction of damage in structures, it is a method that requires the artificial production of vibration or shocks that stimulate the system (or structure or body being monitored). Vibration with unknown multiple magnitudes and frequencies is produced and subsequently transformed during its

transmission through the spine during human gait, thus it is not feasible to employ the algorithms used in structural health monitoring for daily life physical activities.

2.4.3.1. Cutaneous measurement

Transmissibility can be measured with skin-mounted sensors but it is necessary to remove the effect of the skin-sensor interface movement. This correction can be done mathematically by determining a correction factor that will belong to the mounting site of the sensor (Hinz et al., 1988, Kitazaki and Griffin, 1995).

The effect of soft tissue on the measurement of stress wave propagation has been carefully studied and validated previously (Saha and Lakes, 1977, Ziegert and Lewis, 1979, Nokes et al., 1984, Smeathers, 1989b, Trujillo and Busby, 1990, Kim et al., 1993, Kitazaki and Griffin, 1995, Lafortune et al., 1995, Forner-Cordero et al., 2008, Pankoke et al., 2001, Hinz et al., 1988). Saha and Lakes (1977) evaluated the effects of soft tissue on the measurement of vibration produced through a hammer impact on cadaveric human bones and in-vivo on the tibiae of volunteers. Three different accelerometer attachments were tested: static load (in-vivo), spring load (in-vivo) and on a screw inserted directly to bone (in-vitro). Saha and Lakes (1977) concluded that a better correction method than accelerometer loading was necessary to reduce the effect of soft tissue. Ziegert and Lewis (1979), Nokes et al. (1984) and Lafortune et al. (1995) reached a similar conclusion when comparing the simultaneous measurement of an accelerometer connected to the tibia via a needle with a preloaded accelerometer with an elastic strap over the skin. Ziegert and Lewis (1979) found a specific magnitude of preload for the accelerometer at which tissue effects were negligible. Nokes et al. (1984) performed further tests that gave evidence that there is a specific preload magnitude that would allow the measurement of a mechanical response close to that measured directly over the bone. Lafortune et al. (1995) performed the first measurements of vibration transmitted through the tibia while running, in contrast with previous studies where the input stimuli was produced by hammers. Lafortune et al. (1995) found that preloading the sensor was no longer effective for measuring bone vibration during running. From this study it was evident that it was still necessary to find a better method in order to measure stress wave transmission of the

human body during daily life physical activities. Forner-Cordero et al. (2008) tested various attachment methods including preloaded elastic straps, special holders for the accelerometer attached to the skin with adhesive tape and double sided adhesive tape alone. Also different types of stimuli such as nudge test, heel drop and walking were tested. Their main objective was to find a method to characterize the skin-sensor interface response even though elastic strapping had been previously tested and validated for heel strike tests. Forner-Cordero et al. (2008) further confirmed that the magnitude of the load pressing the accelerometer against the skin as well as the weight of the attachment determine the natural frequency of the system, which could limit the kinematic analysis. Yet the attachment methods tested are only applicable to upper and lower limbs and not the spine.

The use of preloaded accelerometers over the skin of human limbs is widely used and effective for kinematic analysis. For the study of the human spine the use of preloaded sensors would be difficult and impractical since the thorax transversal area is different along the spine and it has little resemblance to a cylindrical volume as the legs or arms do. This means that lengthy and uncomfortable tight elastic straps would need to be worn by subjects when performing daily life physical activities in order to hold accelerometers in place over the spinous processes of the spine.

Hinz et al. (1988) described a mathematical method, first described by Artmann et al. (1976), to approximate the skin response and measured RMSa over the spinous processes of T5 and L3 with miniature accelerometers attached to the skin of human subjects with an epoxide compound. Hinz et al. (1988) confirmed previous suggestions that it is necessary to determine the skin response for each individual subject. Hinz et al. (1988) also acknowledged that the skin-sensor interface has nonlinear properties thus the measurements obtained through the skin correction model should be considered an approximation only. Yet it is a reliable method as it was shown that the root mean square acceleration values obtained were well in agreement with previously published values by Panjabi et al. (1986). Kim et al. (1993) further confirmed that the relation between the output of a skin mounted accelerometer and the actual bone acceleration can be successfully represented by a

linear model with a spring and damper. The acceleration corrected for skin movement was compared (in-vitro) with the acceleration of a bone mounted accelerometer and found that the model is valid while the frequencies studied are below the natural frequency of the skin-sensor interface. Kitazaki and Griffin (1995) performed a study with the clear objective of establishing a skin-sensor interface movement correction for measurement of stress wave transmission through the human spine. Their study validated the use of the correction method with accelerometers of different masses. Kitazaki and Griffin (1995) also clarified that the skin correction method needs to be applied individually and for each location over the spine since a standardized correction frequency function is not possible due to the variability between locations and between individuals. Finally Kitazaki and Griffin (1995) also confirmed the negligible nature of skin-sensor interface movement in the anteroposterior direction below 35 Hz. The representation of the skin-sensor interface as a single degree of freedom system was further validated numerically by Pankoke et al. (2001). Although a biomechanical model of a single seated adult male was used, the vertical transmissibility to L4 corrected for skin movement showed better agreement with their model than the one estimated for the uncorrected condition.

The correction for the effect of the skin-sensor interface movement is mainly done in the vertical direction (Y axis). Previous research has shown that the skin response in the mediolateral and anteroposterior directions (Z and X axis) is not significant for the measurement of the transmissibility of the spine (Rubin et al., 2003). Rubin et al. (2003) determined that the transmissibility of low magnitude, high frequency vibration through the horizontal and anterior-posterior directions was less than 10% of that measured at the vertical axis. The skin-sensor interface is assumed to be a local single degree of freedom (SDOF), linear system in order to characterize its movement in the vertical direction (Figure 2.3-2).

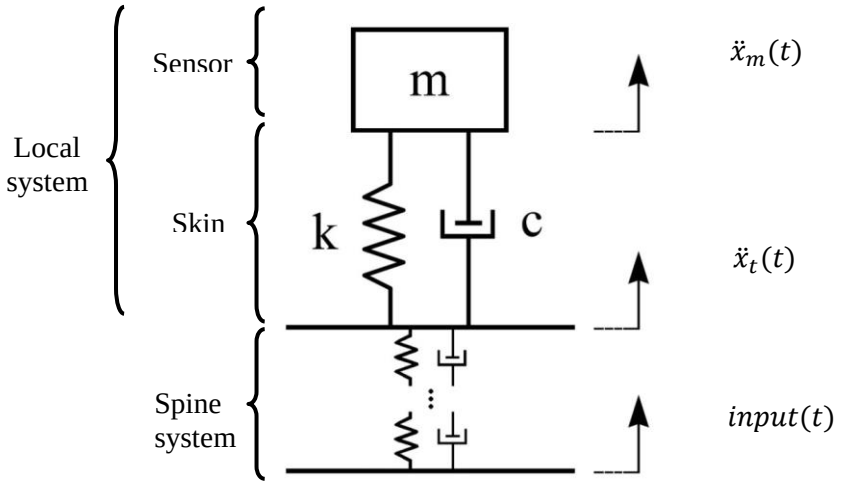

Figure 2.4-2 Single degree of freedom local system representing the skin-sensor interface

The equation of motion by which this system is represented is as follows:

$$m\ddot{x}_m(t) + c\big(\dot{x}_m(t) - \dot{x}_t(t)\big) + k\big(x_m(t) - x_t(t)\big) = 0 \qquad \textbf{2.4-7}$$

Where m is the mass of the sensor and tissue involved in the local vibration, $\ddot{x}_m(t)$ is the acceleration measured at the skin, $\ddot{x}_t(t)$ is the true acceleration of the spine, k is the spring rate of the skin and c is damping coefficient of the skin. Velocity is represented by $\dot{x}$ and displacement by x. A transfer function can be defined between the true acceleration as input and the measured acceleration as output, therefore the response of the SDOF model to a free vibration test can be represented by an equation. A nudge test induces a free vibration response by displacing the skin approximately one centimetre in the vertical direction, followed by a quick perpendicular release of the finger performing the test (Kitazaki and Griffin, 1995). The undamped natural frequency (f_d) and logarithmic decrement (δ) can be calculated from the free vibration response (Figure 2.3-3). The undamped natural frequency (f_d) is given by Equation 2.4-8 and the logarithmic decrement (δ) obtained through Equation 2.4-9. The damping ratio (ζ) is determined from the logarithmic decrement previously calculated (Equation 2.4-10). The natural frequency (f_n) of the skin-sensor system is obtained from the damping ratio and undamped natural frequency previously calculated (Equation 2.4-11).

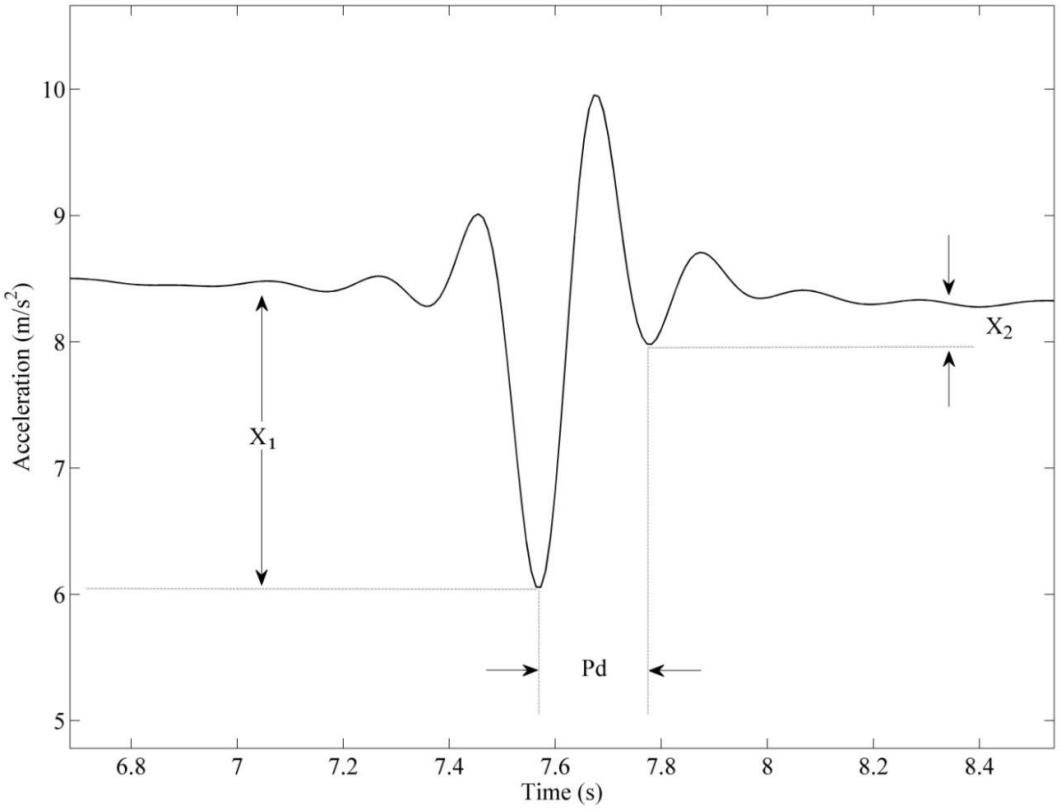

Figure 2.4-3 Typical free vibration response of a skin-sensor interface

$$f_d = \frac{1}{Pd} \quad \textbf{2.4-8}$$

$$\delta = \ln \frac{X_1}{X_2} \quad \textbf{2.4-9}$$

$$\zeta = \frac{\delta}{\sqrt{4\pi^2 + \delta}} \quad \textbf{2.4-10}$$

$$f_n = \frac{f_d}{\sqrt{1 - \zeta^2}} \quad \textbf{2.4-11}$$

A frequency correction function ($C(f)$) of the skin-sensor interface is obtained by substituting the natural frequency and damping ratio previously calculated and a consecutive range of frequencies into Equation 2.4-12, which represents the frequency transfer function between the true acceleration ($\ddot{x}_t(t)$) as input and the measured acceleration as output ($\ddot{x}_m(t)$).

$$C(f) = \left(\frac{1 + \left(2\zeta \left(\frac{f}{f_n} \right) \right)^2}{\left(1 - \left(\frac{f}{f_n} \right)^2 \right)^2 + \left(2\zeta \left(\frac{f}{f_n} \right) \right)^2} \right)^{1/2} \quad \textbf{2.4-12}$$

The true transmissibility of the spine then can be calculated by

$$T_t(f) = T_m(f) \cdot C(f) \quad \textbf{2.4-13}$$

Where $T_t(f)$is the true transmissibility at the spine and $T_m(f)$ is the transmissibility measured at the skin surface (Smeathers, 1989a).

Only a few studies have employed a non-invasive method to study the transmissibility of the human spine during daily life physical activities. Unfortunately, most of them did not employ the skin-sensor interface movement correction method (Voloshin et al., 1981, Voloshin and Wosk, 1982, Wosk and Voloshin, 1981). Since these studies assumed that the effect of the skin-sensor interface movement was negligible during walking, the clinical use of their results is questionable. Similarly, these studies have investigated vertical stress wave propagation from tibial tuberosity and femoral condyle to head mainly via a bite bar, where an accelerometer was attached. Only Wosk and Voloshin (1981) attached a sensor over the sacrum. Wosk and Voloshin (1981) calculated transmissibility as the ratio of peak accelerations rather than in the frequency spectrum through PSD. Nevertheless, their results support the idea that human gait produces vibration that is propagated through the body but fail to characterize it adequately (i.e. skin correction was not performed). Thus the results on attenuation of different sections of the body from Wosk and Voloshin (1981) should be carefully interpreted. In contrast, three studies published in 1989 looked at spine transmissibility of vibration during physical activities such as walking and running while performing the skin-sensor interface correction. Smeathers (1989a) attached accelerometers with adhesive tape over S2 and T2 of two subjects while walking and running. He calculated vertical transmissibility from 1 to 40 Hz and compensated for skin movement with the method outlined previously as well as for the inclination of the accelerometers over the spine. He measured the

inclination of the accelerometers to the vertical (gravity vector) with a goniometer while subjects stood still. He found that the human spine was not as stiff as previously thought and that provided that the skin was not too loose the transmissibility underlying the skin could be measured. This study was the first one to employ stress wave transmission in order to find how the shock absorbing properties of the spine were altered by diseases such as ankylosing spondylitis. Transmissibility results showed that the affected spine behaved as a rigid bar. This study is of paramount importance due to the fact that gives evidence that stress wave propagation during human gait is able to provide information necessary to identify the effects of degeneration on the spine. Helliwell et al. (1989) also measured vibration transmissibility of spines with ankylosing spondylitis and healthy spines. He reached similar conclusions as Smeathers (1989a) while testing walking in more subjects (12), correcting for skin movement and inclination and while measuring transmissibility at a different section of the spine (L4 and T2). One of the major contributions of this study is that it was suggested comparing transmissibility between subjects, i.e. with different health conditions of the spine, using the transmissibility values found at the dominant frequency components of the acceleration measured (corrected for skin movement). This suggestion was made because transmissibility results are difficult to compare since these are comprised of multiple frequencies. The calculation of mean transmissibility may give an overview of the general tendency but it reduces peak values (involves a reduction of information) (Nakai et al., 2007). Smeathers (1989b) attached accelerometers at ten locations over the spine and found that the natural frequency and damping ratio varied along those locations. He also attached an accelerometer on the ankle, sacrum and T2 and measured vertical transmissibility while performing the skin-sensor interface movement correction method outlined previously. He found that legs attenuate most of the frequency components during heel strike. This study confirmed that where the transmissibility of the skin is known, acceleration in the underlying spine can be approximated by accelerations at the skin surface (Smeathers, 1989b).

2.4.4. Summary

Medical imaging combined with computational techniques is still in development to achieve readily available, safe, accurate and validated methods to assess the strength of the osteoporotic human skeleton in clinical practice. In addition, there is a clear tendency in the literature for the use of these specialized medical imaging techniques to explore bone as a tissue and to study the effect of drugs on the bone remodelling process rather than on exploring the extent of physical activity on the prevention and treatment of osteoporosis. Current techniques (DXA, QUS) still fail due to the BMD criteria that cannot identify the structure of bone becoming increasingly weaker. Although high resolution imaging techniques offer a better alternative to BMD in the future, it is uncertain how long it will take before these can be used in clinical practice and how readily available these will be to equally first and third world countries for prevention rather than treatment. On the other hand, low cost techniques such as combined accelerometer and gyroscopes have scope for research through vibration transmission analysis for the characterization of the human skeleton during daily life physical activities. Vibration transmission has the advantage of considering BMD, tissue properties and structure at the same time without the need of high resolution images as well as being portable enabling the study of daily life physical activities.

2.5. General summary

The human spine is a complex structure with different tissues, elements with different geometries and each with a particular function. When spinal osteoporosis is severe and vertebral fractures are present, generally pharmacological and surgical interventions are used even though these are not sufficiently safe and efficient. Non pharmacological interventions need to be further developed. The efficacy and safety of WBV is unknown. Some physical activities have been identified to either produce a modest improvement of spinal BMD or just preserve it. There is the need to find a technique to objectively measure physical activity across individuals with different capabilities and health in vivo.

In vitro biomechanical analysis of the spine has been favoured for the investigation of material properties such as bone and cartilage, restricting in this way the degrees of freedom and potentially

biasing the results due to muscles removal and other soft tissue surrounding the spine. There is scope for research in vivo of the spine, in particular non-invasive study of the osteoporotic spine during daily life physical activities.

Evaluation of the spine in vivo through BMD does not fully account for the structural changes due to osteoporosis, soft tissue properties and for the effect of daily life activities. Although high resolution imaging techniques offer a better alternative to BMD in the future, it is uncertain how long it will take before these can be used in clinical practice. Low cost accelerometers and gyroscopes offer an opportunity to characterize the spinal mechanical environment during daily life physical activities through vibration transmissibility analysis. Vibration transmissibility has the advantage of considering BMD, tissue properties and structure at the same time.

Mechanical stimulation is capable of enhancing the response of bone cells but the biochemical paths of this response are not yet well understood. Vibration produced during human gait offers a non-invasive and non-pharmacological alternative for the treatment of osteopenia and osteoporosis. However, it is necessary to find a way to objectively measure the vibration signals produced by physical activity. Vibration transmissibility analysis may be able to characterize osteoporotic bone's mechanical environment during daily life physical activities. In the future, this technique will enable further research to determine the optimal exercise prescription (safe and efficient) according to individual capabilities and current health status.

2.6. Need for the study

Spinal fractures due to osteoporosis cause disability. Drug treatments in combination with a prescribed diet and exercise moderately improve spinal bone mineral content and strength. It is widely recognized in the literature that there is the opportunity to determine an optimal and safe exercise prescription in order to compensate for what current osteoporosis treatments (including WBV) cannot achieve. Overall, there are apparent bone structural benefits that can be gained exclusively through physical activity. However, there is no technique available to characterize the effects of exercise on the mechanical response of the spine. Without this technique, it is not

possible to characterize the effects of physical activity objectively and reliably across subjects with different BMD and different ages. The technique of vibration transmissibility measurement, being portable and being capable of accounting not only for BMD but for the overall spinal strength, in vivo and noninvasively, needs to be further developed. The feasibility of using vibration transmissibility to detect different physical activities needs to be tested. Similarly this technique needs to be tested in a sample of the population which would include people with osteoporosis.

CHAPTER 3 Vibration Transmission Pilot Study

This chapter is part of a study that has been published as a technical note (Morgado Ramírez et al., 2013c). Further details are presented here.

3.1. Introduction

Vibration is an oscillatory motion with magnitude and repetition rate. The magnitude of vibration can be measured in terms of acceleration and the frequency content through spectral analysis. The simplest type of vibration is a sine wave (Griffin, 1990), but due to the characteristics of human gait, the human body is exposed to complex waves rather than simple sine waves. During gait, each impact of the heel with the ground produces transient waves that propagate up throughout the body (Collins and Whittle, 1989, Cappozzo, 1982). Transmitted waves or vibration may be attenuated and amplified due to muscular contraction (Feltham et al., 2006, Huang and Griffin, 2006) and due to the intrinsic mechanical properties of the tissue through which this vibration is transferred. Accordingly, the shape of these stress waves is expected to change in the time spectrum as they are transmitted (Collins and Whittle, 1989). Transmitted waves up the lower limb during heel strike have not only vertical but also transverse components (Collins and Whittle, 1989). Through the movement of joints during gait, the vertical and transverse vibration travelling through the lower limbs is stored (resembling a spring) and dissipated (resembling a dashpot) (Collins and Whittle, 1989). The attenuation capacity of lower limbs has been investigated previously (Chu et al., 1986, Angeloni et al., 1994). The human spine has multiple elements (bone, muscle, ligaments, tendons, cartilage) which may store, generate and dissipate energy (Voloshin and Wosk, 1982). It has been suggested that the spinal column has shock absorbing properties when subjected to mechanical shocks (Sandover, 1988), whole body vibration devices (Sandover, 1988) and while walking and running (Smeathers, 1989a). However, it may have different responses at different frequencies when exposed to random and complex waves produced during daily life activities. Assessment of

the mechanical stimuli associated with daily life activities will develop our understanding on the effects of these activities on the musculoskeletal system.

The dynamic properties of the spine can be expressed in the frequency spectrum as the transmissibility of vibration from the sacrum to the upper end of the thoracic spine (Mansfield, 2005b). This vibration can be measured with accelerometers attached to the skin over bony prominences after performing corrections for the skin-sensor interface movement (Kitazaki and Griffin, 1995) and for the inclination of the sensor in relation to the vertical (Smeathers, 1989a). This method has been previously validated against pins inserted directly to bone (Kitazaki and Griffin, 1995, Kim et al., 1993, Pankoke et al., 2001). Currently there is no strict guideline for the use of specific brands and models of accelerometers for the measurement of vertical human vibration transmissibility. Therefore the feasibility of performing this measurement has to be carefully tested with customized correction procedures (skin-sensor interface movement and sensor inclination).

The transmissibility capacity of the human spine has been studied previously during walking (Smeathers, 1989a), running (Smeathers, 1989a) and heel strike (Smeathers, 1989b), but not extensively during stair negotiation. Smeathers (1989a) attached one accelerometer over the second sacral vertebra and another over the second thoracic vertebra with adhesive tape. He calculated vibration transmitted through the spine by correcting for skin movement and sensor inclination in the sagittal plane using the accelerometers as inclinometers before subjects performed walking and running. The main limitation of this study was that a constant inclination angle of the trunk was assumed in determining transmissibility, although the spine orientation may change significantly during these activities (Crosbie et al., 1997). Moreover, the description of vibration transmissibility measurements during daily life physical activities has not been performed rigorously but only through visual analysis of the vibration patterns.

The present study addresses the limitations of previous work and extends the transmissibility analysis to a wider range of daily activities. The purpose of this study was (1) to examine the feasibility of measuring the transmission of vibration through the human spine using skin mounted

inertial sensors and, (2) to assess the dynamic properties of the spine during activities of daily living. The effect of corrections will be objectively established. The measurement method will be employed to determine transmissibility through the spine during level ground walking along a straight line and during stairs ascent and descent.

3.2. Methods

Ten young and healthy participants were recruited; individual details can be seen in Table 3.2-1. Subjects were excluded if they had experienced back or leg pain in the last 12 months that required medical treatment, rheumatological disorders, dislocation, fracture or surgery of the spine or lower limbs, neurological disorders which affected their gait and if they were obese (with a body mass index greater than 29 kg/m^2). All subjects underwent a Broad Ultrasound Attenuation test to determine their bone mineral density. This test was performed using a Quantitative Ultrasound Scanning (QUS) system (CUBAClinical, McCue Plc.) with dedicated software (CUBA Plus, McCue Plc.). Left and right heel bones (calcaneus) were tested in order to identify the heel with the lowest T-score. QUS results expressed in terms of the T-score and World Health Organization guidelines (WHO, 1994) were used to select only those subjects with normal density (T-score > -1.0) and in their peak bone mass (Figure 2.1-2). As mentioned before, BUA has been shown to assess osteoporosis by both prospective and retrospective studies (Bauer et al., 1997, Bauer et al., 1995). Subjects were excluded if they had osteopenia or osteoporosis (T-score < -1). Selection of healthy subjects was necessary in order to reduce biased results since the vibration transmissibility measurement is dependent on geometry (vertebrae) and material properties (soft tissue and bone). Ethical Approval was given by University of Roehampton ethics committee (Appendix A). All volunteers gave informed consent by signing the approved consent form (Appendix A). Subjects were asked to wear loose clothing and the shoes that they used most of the time excluding high heels and sandals. A disposable gown was provided to each participant to wear during the sensor attachment process. This enabled access to the sensors without exposing the volunteer's body.

Three inertial sensors (Wireless InertiaCube3™, InterSens Inc.) were put over three locations of the spine. Each inertial sensor had a weight of 20 g, an operating range of ± 2 g and comprised of three

dimensional accelerometers and gyroscopes, which were used to measure vertical acceleration and angular rotation of the sensor respectively. To evaluate the suitability of the inertial sensors for vibration transmissibility measurement, two inertial sensors were put side by side over the first thoracic spinous process (T1) allowing for each to move in their vertical direction without touching each other. These sensors recorded the output signals on two sides of the T1 vertebra. Each sensor being in a different location had a different source of error since the skin properties are diverse in different parts of the spine (Smeathers, 1989b, M-Pranesh et al., 2010). Thus the assessment of the similarity of these two output signals after signal correction helped to establish the effectiveness of the correction procedure. Another accelerometer was put over the first sacral vertebra (S1) to measure the input signals (Figure 3.2-1). All sensors were aligned in the sagittal plane of the spine (to be able to measure vertical acceleration) and attached to the subject's skin with double sided adhesive tape. The accelerometers detected linear acceleration while the gyroscopes sensed angular rotation.

Table 3.2-1 Pilot study subjects details, individual and mean (SD)

Subject	Gender	Height (m)	Mass (kg)	BMI (kg/m^2)	T-score	Age
1	M	1.71	64.8	22.16	1.543	28
2	M	1.7	65	22.49	-0.237	27
3	M	1.64	54.95	20.43	-0.343	25
4	M	1.86	71.8	20.71	-0.807	32
5	M	1.74	66.75	21.92	-0.873	25
6	M	1.85	76	22.21	-0.227	25
7	F	1.55	53.3	21.96	-0.407	36
8	M	1.77	76.4	24.25	1.797	33
9	F	1.67	62.1	22.27	0.220	25
10	F	1.63	58.5	22.02	0.373	27
Mean (SD)		**1.71 (0.09)**	**64.96 (8.08)**	**22.04 (1.03)**	**0.104 (0.91)**	**28.3 (3.97)**

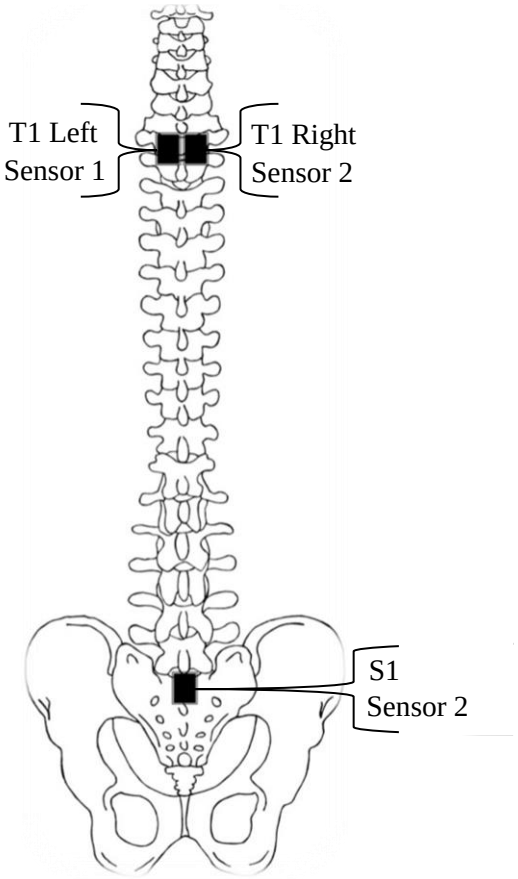

Figure 3.2-1 Location of inertial sensors for vibration transmissibility pilot study. First thoracic vertebra (T1), first sacral vertebra (S1)

To correct for skin movement, all skin-sensor interfaces were subjected to "nudge" tests. This test assumes that the skin has a linear response thus it is represented by a single degree of freedom system (Kitazaki and Griffin, 1995). The nudge test was repeated four times on each sensor to provide an estimate of the correction parameters. During these tests subjects were asked to stand still and to look forward with arms by their sides in a relaxed and comfortable way. The test involved manual displacement of the skin above the sensor by approximately one centimetre in the vertical direction, followed by a quick release of the finger performing the test (Kitazaki and Griffin, 1995). All nudge tests were performed with the same finger and by the same investigator. Acceleration measured during this test leads to a free vibration response of the skin-sensor system, which allows the calculation of a damping factor (ζ) and natural frequency (f_n) (Kitazaki and

Griffin, 1995). These values are employed in the skin-sensor interface movement correction method which has been validated elsewhere (Kim et al., 1993, Ziegert and Lewis, 1978).

Subjects were then asked to perform three activities three times at a self selected, normal speed (NWS): walk in a straight line (33 m in length and 2 m wide), ascend and descend standard stairs consisting of 15 steps of normal height and 1.19 m wide with a continuous hand rail on both sides (Figure 3.2-2). Vertical acceleration and dynamic sensor inclination were wirelessly stored in a laptop computer for each inertial sensor. Wireless timing gates (Smartspeed™, Fusion Sport Pty Ltd.) were used to measure the time that each subject took to complete each walking trial. These times were used to calculate average walking speed. A rest was given between trials to prevent fatigue.

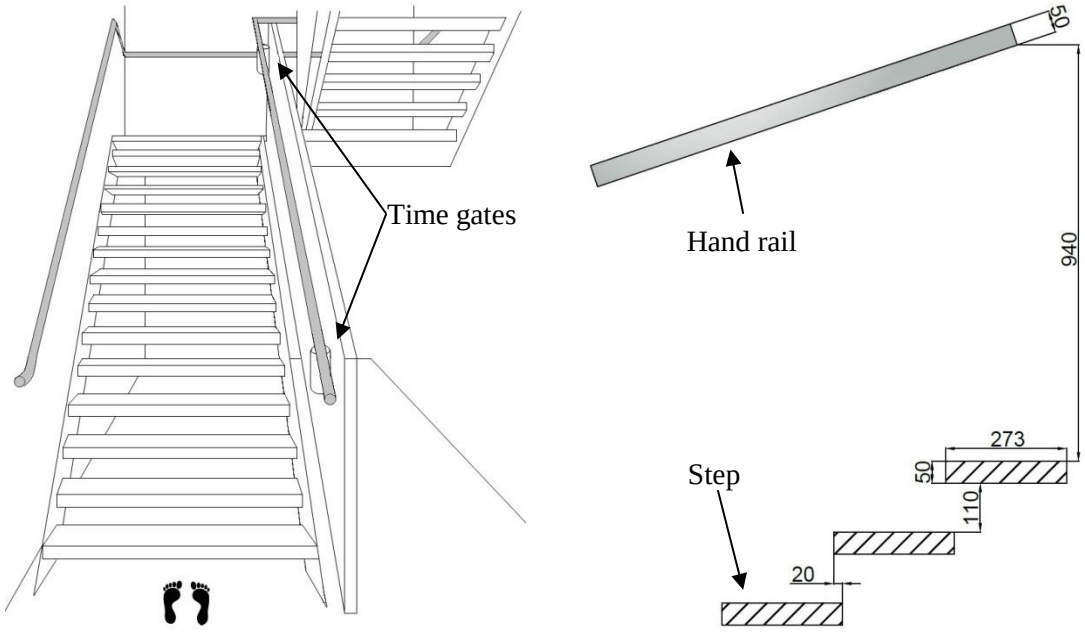

Figure 3.2-2 Staircase with time gates on place and subject feet in the start position. Lateral view of a section with dimensions in mm

3.3. Data processing and analysis

Data was analysed using custom made scripts in Matlab (The Mathworks Inc., 2010) and with SPSS statistical software (SPSS Inc., 2009). Raw sensor acceleration and angular rate signals were unevenly sampled by the software that the manufacturer provides for the inertial sensors (IsPlot 1.006, InterSens Inc.). Thus these signals were resampled at 110 Hz as well as low pass filtered at

20 Hz with a zero phase 5th order Butterworth algorithm. A maximum of 20 Hz was selected because human gait frequency content was previously reported (Cappozzo, 1982, Angeloni et al., 1994) to lie between 0.75 and 9.8 Hz. This study explored transmissibility of vibration up to 12 Hz due to the natural frequency of the skin-sensor interface obtained.

The acceleration signals were subjected to three different corrections before the calculation of transmissibility. The first correction (correction 1) consisted in correcting for the inclination of the sensor when attached to the back of the subjects, the second correction (correction 2) consisted in correcting for the movement of the skin and the third and last correction (correction 3) consisted in removing the effect of gravity on the corrected acceleration.

The angular rate signals produced by the gyroscopes of the inertial sensors were employed to determine θ. Specially written Matlab scripts were used to account for the angle θ so that acceleration signals were corrected with respect to the vertical for each sensor and subject. This angle was identified as pitch by the manufacturer's software. A custom made script in Matlab allowed for the sensor's gyroscopes to report a pitch of 0° when perfectly aligned to the vertical (gravity vector). Thus when attached to the spine the gyroscopes were able to measure the inclination of the inertial sensor to the vertical as a positive angle θ.

Correction 1: signals were corrected for inclination of the sensor in the time spectrum. If θ is the angle to the vertical representing the sensor rotation about the mediolateral axis over the spine at each instant of time (i), and $\mathrm{a_m}$ is the acceleration measured on skin surface, the vertical acceleration ($\mathrm{a_v}$) can be determined as:

$$a_v = \mathrm{a_{m_i}} \cos\theta_i \qquad \textbf{3.3-1}$$

Correction 2: this involved acceleration correction for skin-sensor interface movement in the frequency spectrum (f) employing the frequency correction function $C(f)$ (Equation 2.3-12). The acceleration below the skin ($\mathrm{a_s}$) can be determined as:

$$\mathrm{a_s} = \mathrm{iFFT}(\mathrm{a}_v(f)\, C(f)) \qquad \textbf{3.3-2}$$

where iFFT is the inverse Fourier transform.

Correction 3: Fully corrected acceleration was completed by the subtraction of the arithmetic mean (a_m) in order to include only that vibration unrelated to the earth's gravity (Equation 3.3-3).

$$a_c = a_s - \overline{a_m} \qquad \textbf{3.3-3}$$

Transmissibility of vertical vibration along the spine was estimated as the square root of the ratio of the power spectral density (PSD) of each output (left and right T1 separately) over the PSD of the input (S1) over the frequency interval of 0.5 to 12 Hz (Equation 2.3-4).

A Kruskal-Wallis one way ANOVA test ($p<0.05$) was used to determine if the angle to the vertical was significantly different across the physical activities tested for all sensors. In order to assess the effectiveness of the correction, the signals acquired by the two T1 sensors during the various physical activities were compared through cross correlation in the time spectrum and through coherence in the frequency spectrum (Shin and Hammod, 2008) between the uncorrected and the fully corrected condition. In addition, the reliability of the signals of the three sensors was evaluated by cross correlation between the three trials. Significant differences after correction were calculated through the Mann-Whitney U test.

Transmissibility of the vertical acceleration signals was plotted as a function of the frequency of the signals. Maximum transmissibility values were determined to examine the frequencies at which the highest amplification of vibration was obtained. Kruskal-Wallis one way ANOVA ($p<0.05$) was carried out to examine the differences in the mean maximum transmissibility between the various physical activities for three frequency (f) intervals ($0.5\leq f<4$, $4\leq f<8$, $8\leq f<12$ Hz).

Mean maximum acceleration spectral density was determined at S1 and left and right T1 after full correction. Significant differences in the mean maximum spectral densities between physical activities were determined through a Kruskal-Wallis one way ANOVA test ($p<0.05$) for three frequency intervals ($0.5\leq f<4$, $4\leq f<8$, $8\leq f<12$ Hz). Non-parametric tests were employed because of the small sample size.

3.4. Results

Mean self-selected and normal walking speeds were 1.59 ± 0.29 m/s for straight level ground walking, 0.56 ± 0.06 m/s for ascending stairs and 0.64 ± 0.09 m/s for descending stairs.

Mean natural frequencies of the skin-sensor interfaces were between 11.7 and 17.4 Hz (Table 3.5-1). The lowest natural frequency was found for the skin-sensor interface of the left side of T1 while the highest was found for the right side of T1. The values of ζ and f_n are different between left and right sides of T1, likewise the frequency correction functions employed.

While walking, the left and right sensors over T1 mean angles to the vertical were different and had a magnitude of 55 ± 8.2° and 56.1 ± 6.7° respectively. These mean inclination angles were different while ascending (left: 45.5 ± 6.5°, right: 49.4 ± 6.2°) and descending stairs (left: 51.4 ± 5.7°, right: 54.1 ± 5.1°) but still presented standard deviations greater than 5.1°. The sensor over S1 had mean angles to the vertical of 77.8 ± 6.9°, 69.6 ± 8.9° and 78.1 ± 6.5° while walking, ascending and descending stairs correspondingly. The angle to the vertical was significantly different across the physical activities tested for all sensors.

3.4.1. Correction of the acceleration signals

The fully corrected vertical accelerations obtained during level ground walking and their frequency spectrums for one of the subjects are shown in Figure 3.4-1 as a typical example. The patterns of the corrected acceleration signal at both outputs changed compared with the input. The four harmonic frequencies of the input were attenuated by approximately half or less of their magnitude. Main frequency components transmitted through the spine were observed in all the frequency intervals studied (Figure 3.4-1).

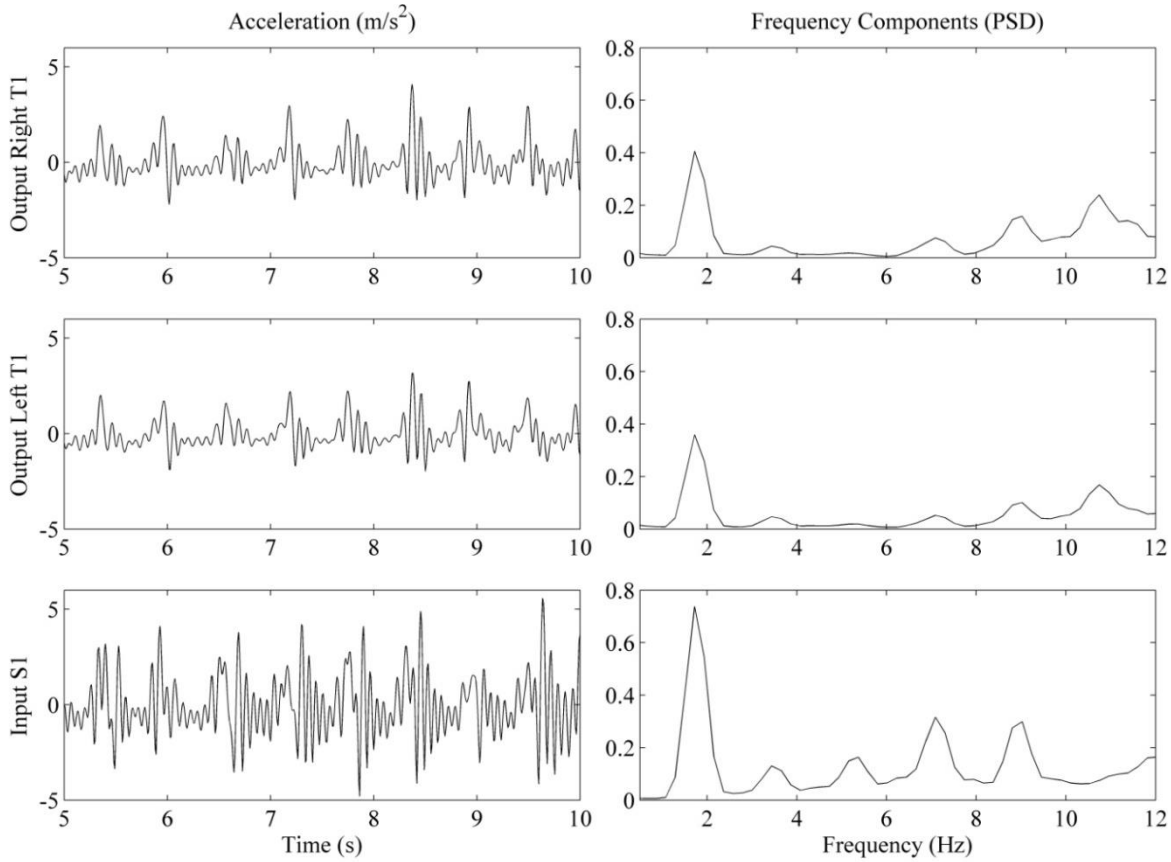

Figure 3.4-1 Fully corrected vertical acceleration at input (S1) and outputs (T1) and their frequency spectrum during level ground walking. First sacral vertebra (S1), first thoracic vertebra (T1)

Cross-correlation of the two T1 fully corrected signals was high (>0.9) for all three physical activities. The cross-correlation coefficients increased significantly after full correction for walking (Table 3.4-1). Correction for sensor orientation alone did not improve the cross-correlation of the signals significantly. Cross amplitude spectrum also showed maximum association between the two T1 fully corrected signals at approximately 2 Hz for all physical activities (Table 3.4-1). Mean maximum coherence at the frequency where mean maximum amplitude resulted in strong associations (>0.9) (Table 3.4-1).

The inter-trial cross correlations of all three sensors were high (>0.9), showing that the signals were consistent among trials (Table 3.4-2). All cross-correlation coefficients increased after full signal correction except for S1 while walking and ascending stairs. These also increased significantly for

all sensors while ascending stairs and for S1 while descending stairs (Table 3.4-2). Walking showed no statistically significant difference after correction at all locations. There was no significant improvement of cross correlation of acceleration at left and right side over T1 after correction while descending stairs (Table 3.4-2). The sensors were considered to be able to produce repeatable signals for the transmissibility analysis.

Table 3.4-1 Mean cross-correlation coefficients and coherence for inter-sensor comparison (between left and right sides at T1)

Inter-sensor (Left and right side over T1)			
Activity	**Mean Maximum Cross Correlation**		**Mean Maximum Coherence at maximum CPSD amplitude (C)**
	U	**C**	
Straight Walking	0.992 (0.006)	0.985 (0.010)	0.945 (0.046) at 1.987 Hz
Stairs Ascent	0.998 (0.002)	0.997 (0.003)	0.957 (0.067) at 1.826 Hz
Stairs Descent	0.929 (0.334)	0.986 (0.015)	0.924 (0.069) at 2.148 Hz

First thoracic vertebra (T1), cross power spectral density (CPSD), mean (SD), uncorrected (U), corrected (C)

Table 3.4-2 Mean cross-correlation coefficients for inter-trialcomparison of all three sensors (left and right T1 and S1)

Inter-trial Mean Maximum Cross Correlation						
Activity	**Left T1**		**Right T1**		**S1**	
	U	**C**	**U**	**C**	**U**	**C**
Straight Walking	0.964 (0.032)	0.966 (0.022)	0.965 (0.029)	0.967 (0.015)	0.968 (0.015)	0.967 (0.015)
Stairs Ascent	0.992 (0.010)	0.996 (0.005)	0.993 (0.009)	0.997 (0.004)	0.981 (0.030)	0.997 (0.004)
Stairs Descent	0.980 (0.017)	0.987 (0.015)	0.971 (0.040)	0.984 (0.019)	0.964 (0.056)	0.989 (0.011)

First thoracic vertebra (T1), first sacral vertebra (S1), mean (SD), uncorrected (U), corrected (C)

3.4.2. Transmissibility of vertical acceleration during physical activities

Typical examples of transmissibility patterns during the three physical activities tested are shown in Figure 3.4-2. Mean maximum transmissibility values with a 95% confidence interval were determined for three different frequency intervals (Figure 3.4-3). Transmissibility above 1 means

amplification, where 1 is equal to 100% of vibration transmitted. Mean maximum transmissibility values were compared between physical activities; these were dissimilar at different frequency intervals (Figure 3.4-3). Level ground walking at a normal speed, amplified vibration in the frequency interval studied with a minimum amplification of 120 ± 33% (4≤ f <8 Hz) and maximum of 134 ± 39% (0.5≤ f <4 Hz). Ascending stairs amplified the input vibration from 8 to12 Hz by 114 ± 34% but attenuated vibration in other frequency intervals (below 73 ± 19% of transmission).

Descending stairs attenuated vibration signals (transmissibility of less than 92 ± 30%) over the entire frequency interval studied. Mean maximum transmissibility was found to be significantly different between walking and descending stairs for the frequency interval studied (Figure 3.4-3). Transmissibility while ascending and descending stairs was significantly different only for the frequency interval 4≤ *f* <12 Hz. Transmissibility was not significantly different while walking and ascending stairs for the frequency interval 4≤ *f* <12 Hz.

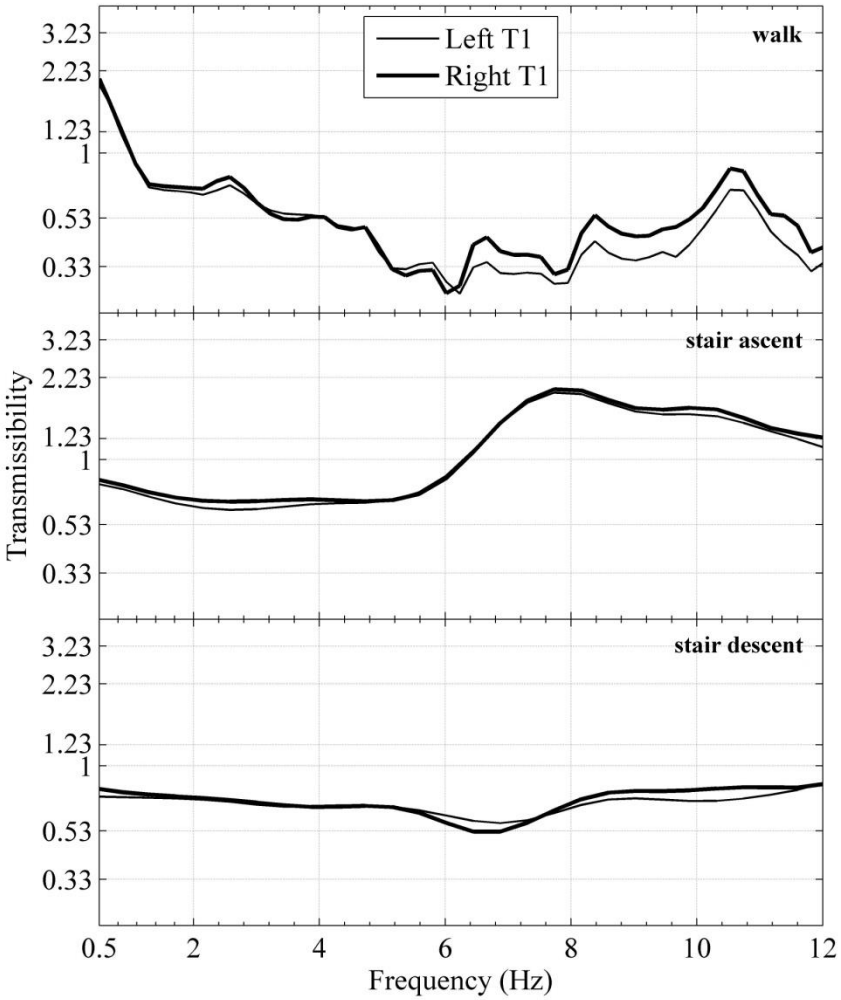

Figure 3.4-2 Typical transmissibility responses on the two sides of the first thoracic vertebra (T1) while straight walking, ascending and descending stairs for one subject

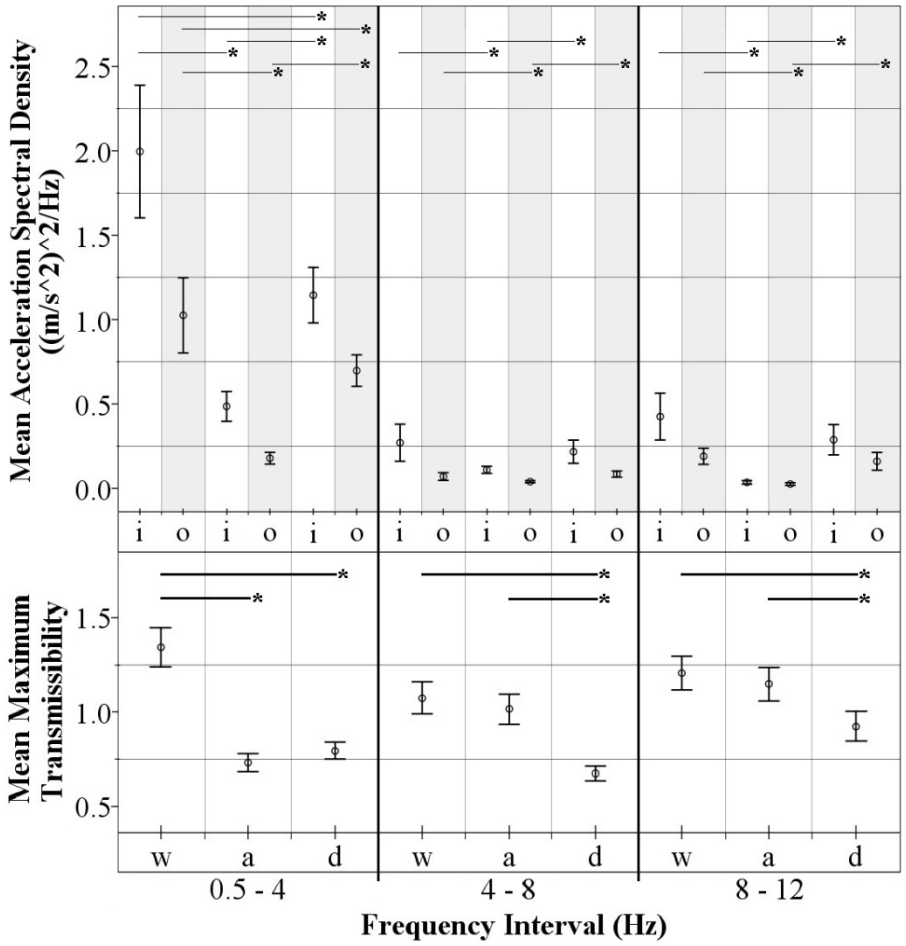

Figure 3.4-3 Mean maximum transmissibility and mean maximum spectral density of fully corrected acceleration at input (i) and outputs (o); for all physical activities at three frequency intervals (0.5 ≤ f < 4, 4 ≤ f < 8, 8 ≤ f < 12 Hz) with 95% confidence interval error bars. * = significant difference between physical activities

3.4.3. Spectral density of vibration

Mean maximum acceleration spectral density at S1 during walking was found to be significantly greater when compared with ascending stairs over the entire frequency interval studied (Figure 3.4-3). Similarly, the spectral density of vibration at S1 was significantly greater during descending stairs when compared with ascending stairs over the whole frequency interval studied. For the 0.5≤ *f* <4 Hz frequency interval acceleration spectral density was significantly greater while walking in comparison with descending stairs. After this vibration was transmitted through the spine and measured at the level of T1, the mean maximum acceleration spectral density was still significantly

different between walking and ascending stairs and between ascending and descending stairs. The spectral density of vibration was also significantly greater while walking in comparison with descending stairs over the frequency interval of $0.5 \leq f < 4$ Hz. Acceleration spectral density at T1 was not significantly different between walking and descending stairs from 4 to 12 Hz. Acceleration spectral density was found to be less than 0.75 $(m/s^2)^2/Hz$ between 4 and 12 Hz.

3.5. Discussion

This study examined the effects of correction of vibration signals using the characteristics of the skin-sensor interface and the sensor orientation. Such correction enables the inference of the transmission of signals along the spine. Previous work examined the correction method in walking and running only (Smeathers, 1989a) but in the present study the effectiveness of correction during ascending and descending stairs was also examined.

Inertial sensors which comprised accelerometers as well as gyroscopes were employed in this study. Sensor inclination was found to change significantly during physical activities, and thus it would be important to correct the signals using the inclination information. However, in previous work, the inclination was assumed to be constant (Smeathers, 1989a, Hinz et al., 1988). This was because only accelerometers were used and no sensor inclination data was available during physical activity. It is recommended that full correction should be used because the inclination and curvature of the spine (which would significantly influence sensor orientation) could vary significantly in different subjects (for example, older people (Singh et al., 2010)) and during different physical activities (Crosbie et al., 1997). However, it should be noted that although the inclusion of the gyroscope allows the correction to be performed, it does increase the weight of the sensor thus decreasing the natural frequency of the skin-sensor interface. In this study the natural frequencies of the skin-sensor interfaces responses allowed the study the vibration produced by daily life physical activities.

To assess the effect of each skin-sensor interface, the damping ratio and natural frequency were first determined. Left and right sensors did not measure the same error source. Results showed that the correction method employed enabled the detection of a strong association of the vibration

below the skin, particularly at approximately 2 Hz. Mean values of ζ and f_n reported in this study are compared with previously published data (Table 3.5-1). There are significant differences among the various studies in regard to the sensor attachment method, the weight of the sensor and the locations of the spine studied. These factors, together with differences in the mechanical characteristics of the soft tissues, contribute to the differences in ζ and f_n reported. Corrections for the skin-sensor interface movement were performed for each subject and for each skin-sensor interface, mean ζ and f_n were reported for comparison with previous studies only.

Table 3.5-1 Comparison of natural frequency and damping of the skin-sensor interface with previous studies

	This study			**Kitazaki and Griffin (1995)**		**Smeathers (1989a)**		**Hinz et al. (1988)**			
Location	T1 Left	T1 Right	S1	L3	L2•	T2	S2	T1	T5	L3	S1
f_n **(Hz)**	13.8 (2.1)	14.3 (3.0)	14.1 (2.2)	23.6 (6.2)	10.2 (2.2)	19 (—)	12.5 (—)	8.6 (—)	5.9 (2.1)	11.0 (19.7)	11.4 (—)
ζ	0.15 (0.06)	0.16 (0.03)	0.13 (0.05)	0.29 (0.05)	0.30 (0.08)	0.39 (—)	0.26 (—)	0.18 (—)	0.013 (0.003)	0.29 (0.05)	0.44 (—)
Sample size	10			8		1		1			
Sensor weight (g)	20			25.4		3.4		0.5			
Attachment technique and total surface contact area	33.7 mm by 26.1 mm and double sided adhesive tape			35 mm by 40 mm stiff card and double sided adhesive tape		60 mm^2 of adhesive tape over the sensor an skin		Epoxide compound, contact area not reported.			

Natural frequency (f_n), damping ratio (ζ), first thoracic vertebra (T1), second thoracic vertebra (T2), fifth thoracic vertebra (T5), first sacral vertebra (S1), second lumbar vertebra (L2), third lumbar vertebra (L3)

By comparing the corrected acceleration signals at left and right sides of T1 using cross-correlation (time domain) and coherence (frequency domain), a strong association between corrected signals was found. This suggests that the current protocol provided reliable information about the signals that are transmitted through the spine. However, it is acknowledged that cross correlation may not be able to completely reveal the complexity of improvements after correction. It would be necessary to compare the corrected signals with those from sensors attached to pins which were

surgically inserted into the spinous processes (Rubin et al., 2003). However this would be unethical and impractical. In vitro experiments to compare vibration measured with pins and skin mounted sensors have been performed previously (Ziegert and Lewis, 1978, Kim et al., 1993). Kim et al. (1993) used two identical accelerometers, one mounted on a lag screw inserted into the amputated tibia of a human leg and the other skin mounted through a T-shaped aluminium adapter and strapped with a rubber band. The aluminium support added weight to the accelerometer but the band laid a compression load, diminishing the effect of the extra weight. A pendulum impactor of 2 kg simulated the heel strike. Kim et al. (1993) concluded that the skin correction method alone was sufficient to detect the signals resulting from heel strikes. While in this study there is no additional adapter for the sensor (card or T-shaped adapter) or an elastic interval laying a compressive load, the natural frequencies of the skin-sensor interfaces responses allowed the study of the vibration produced by daily life physical activities.

Previous studies (Smeathers, 1989a, Kitazaki and Griffin, 1995) have not quantitatively demonstrated the improvement in signals after correction for skin movement and sensor inclination. The cross-correlation between the left and right T1 sensors showed definite improvement in signals, although the correction for spine orientation alone does not appear to increase the cross-correlation. Walking and descending stairs suggests that there is no significant improvement in the acceleration signal after correction yet this result might not be statistically significant due to the sample size employed. Moreover, spine vibration transmissibility was more reliable after full correction. All the above evidence supports that the correction algorithm is effective. The results of this study suggest that full correction should be used at all times because the inclination and curvature of the spine (which would significantly influence sensor orientation) could be very diverse in different subjects (Singh et al., 2010) and during different physical activities (Crosbie et al., 1997).

Vibration transmissibility patterns were different between dissimilar physical activities. For all the activities tested, the spine transmitted more than 50% of the vibration received at the sacrum bone.

The physical activities tested showed different mean maximum transmissibility values which were found to occur at different frequencies. This means that different physical activities have dissimilar frequency characteristics and as a result amplification and attenuation of vibration takes place at particular frequency intervals. Level ground walking was the only physical activity that amplified the vibration at the lower frequencies ($0.5 \leq f < 4$ Hz) exceeding 125%. Mean maximum transmissibility was rather varied throughout the other frequency intervals. The interpretation of signals above 12 Hz is difficult because these would be attenuated by the skin. This is not a matter of concern as the frequency spectra of most of the activities tested was seen between 0.5 and 12 Hz, and the current data shows that the spectral density of the signals was small at higher frequencies.

One limitation of the present study is that lateral inclination of the inertial sensors was not taken into account, but sensors were carefully attached in the sagittal plane. Gyroscopic data is useful for sensor orientation correction, but needs to be adjusted for drift using information from magnetometers as in the case of the inertial sensors used in this study. Thus the sensors must be used away from metal structures. This would limit the use of the technique where there is metallic interference. The use of full gyroscopic data in clinical settings may be limited, for instance most buildings and staircases have extensive ferrous metal elements which cause interference of the sensor data. Another limitation was the dimension of the inertial sensors employed. The sensor over S1 covered a partial section of the sacral vertebrae below and the sensors over T1 covered only a partial lateral section of it. Kitazaki and Griffin (1995) assumed that increasing the contact area of the sensors over the skin would increase the skin's stiffness and provide stable sensor's motion. Similarly, Kitazaki and Griffin (1995) suggested it would be better to have a surface contact area equivalent to a vertebral body's approximate size but did not discuss the size of the spinous processes. However, previous studies have applied a wide variety of these conditions (Smeathers, 1989a, Kitazaki and Griffin, 1995, Hinz et al., 1988, Ziegert and Lewis, 1978, Kim et al., 1993, Helliwell et al., 1989). Lastly, QUS provides only an approximate T-score as a guidance of the general health of the skeleton, while DXA is preferred for the clinical diagnosis of osteoporosis and drug treatment monitoring.

The present findings suggest that different physical activities may produce mechanical signals with different biomechanical responses as the transmissions of signals are diverse in different activities. It should be noted that in understanding the mechanical effects of the vibration signals, one must consider transmissibility as well as the amplitude of the signals. The biological effects of the transmitted vibration should be established in future studies. It has been previously suggested that it is possible to find a relationship between vibration transmission and pathological changes of the human spine (Helliwell et al., 1989). Finally, Hill et al. (2009) pointed out that current research on vibration, as a stimulus of the human spine, lack a detailed factorial exploration of frequency and amplitude of the signals. This study explored the frequency content and amplitude of vibration transmitted through the young and healthy human spine during daily life physical activities. It is important to understand the harmful and beneficial effects of vibration on the health of the human spine. The measurement technique presented can be used to study signal transmission across the human spine of the older population. It would be useful to study how degeneration of the spine or other disease processes such as osteoporosis affect signal transmission. This study has identified a portable and reliable measurement technique which could be used for such purposes.

3.6. Conclusion

This study employed skin-mounted inertial sensors to study the transmission of signals through the spine. The vibration signals detected by the sensors were corrected for errors due to skin deformation and inertial sensor inclination. Surface measurement of the transmissibility of signals through the spine was found to be accurate and reliable. It is concluded that surface measurement of vertical vibration transmissibility over the spine and during daily life activities is possible with the correction method presented. The results suggest that different physical activities might produce different mechanical stimuli and biomechanical effects on the bone. We recommend that the present measurement protocol be employed for future studies on vibration transmission in older people.

3.7. Key findings

1. It is suggested that the current protocol provides reliable and accurate information about the signals that are transmitted through the spine.
2. Vibration transmissibility patterns were different between dissimilar physical activities.

CHAPTER 4 Vibration Transmission through the Lumbar Spine

4.1. Introduction

In the year 2000, 1.4 million vertebral fractures were related to osteoporosis worldwide (Johnell and Kanis, 2006). Vertebral fractures have a significant impact on daily life activities since they cause back pain, loss of height, deformity, immobility and reduced pulmonary function (IOF, 2012). While there are pharmacological interventions for the prevention and treatment of osteoporosis, these are limited due to their cost, side effects and issues with long term compliance (Hamilton et al., 2010a). Pharmacological treatments increase spinal BMD from 1% to 15% and reduce vertebral fractures risk from 30% to 83% depending on gender and on the drug and time used (NOF, 2010, IOF, 2012, Burr et al., 2002). Physical activity may prevent osteoporosis, and may be used with pharmacological interventions for the management of osteoporosis (IOF, 2012, Burr et al., 2002). Recent reviews suggest that BMD improvements due to physical activity are only modest (less than 2% spinal BMD increment), site specific and have more effect on cortical than trabecular bone in contrast with pharmacological treatments (Hamilton et al., 2010a, Gómez-Cabello et al., 2012). Physical activities that have been identified to either produce a modest improvement of spinal BMD or just preserve it are sparse (walking, volleyball, Tai Chi, aerobics, strength training and a combination of physical activities) (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Cheung and Giangregorio, 2012). Current physical activity measurements do not take into account bone structural changes (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Gremeaux et al., 2012, Cheung and Giangregorio, 2012). Studies attempting to determine if physical activity decreases the degenerative effects of osteoporosis in older adults, often explore changes in metabolic and cardiovascular stress (percent maximal heart rate, percent of one repetition maximum, maximal oxygen consumption) or changes in BMD (measured through X-ray absorptiometry) and occasionally bone structure (measured with

peripheral quantitative computed tomography). However, there is a lack of information about how vibration is transmitted through the lumbar spine during physical activities. This may be useful for understanding the effects of exercise and osteoporosis on the lumbar spine. It is necessary to employ a pragmatic way to characterize the effect of physical activities on bone noninvasively, especially for the older population (Gómez-Cabello et al., 2012, Gremeaux et al., 2012).

Extensive research has led to the understanding that the processes of bone formation and resorption are responsive to mechanical factors (Skerry, 2008, Chen et al., 2010). Bone responds to mechanical stimulation in the form of vibration and the way this vibration is transmitted through the bone will depend on its material and structural properties (Keller et al., 2000, Bediz et al., 2010, Bhattacharya et al., 2010, Kawchuk et al., 2009). Heel strikes during gait produce vibration that is transmitted through the body (Collins and Whittle, 1989, Cappozzo, 1982). Two measures for analysing the nature of the mechanical stimulation on the bone are vibration transmissibility and vibration magnitude. The measurement of vibration transmitted through bone in vivo offers an option to objectively measure the effects of different physical activities on individuals of all ages and all bone health status. This technique consists in measuring the vibration transmitted through the human body and produced during gait (Morgado Ramírez et al., 2013c). Transmissibility greater than 100% indicates amplification while attenuation is indicated by less than 100% (Mansfield, 2005a). Inertial sensors are attached to the spine with adhesive tape and the movement of the skin where the sensor is attached corrected as reported previously (Saha and Lakes, 1977, Ziegert and Lewis, 1979, Hinz et al., 1988, Smeathers, 1989a, Smeathers, 1989b, Kim et al., 1993, Kitazaki and Griffin, 1995, Pankoke et al., 2001). Vibration transmissibility through the spine has been measured previously during walking and running in only two young and healthy subjects (Smeathers, 1989a, Smeathers, 1989b). However, the feasibility of using vibration transmissibility to identify the effect of ageing and osteoporosis on the lumbar spine has not been explored. The magnitude of the vibration transmitted to the spine can be presented in terms of root mean square acceleration (RMSa) which is equivalent to the standard deviation of acceleration produced during gait (Mansfield, 2005a). This is a single value for all frequencies expressed in m/s^2. Previous

studies have investigated the effect of mechanical stress on bones and presented results in terms of micro strains (με), Newtons (N) and acceleration (g forces and m/s^2) related to changes in either BMD or biochemical markers of bone metabolism (Vainionpää et al., 2009, Asselin et al., 2011, Al Nazer et al., 2012, Burr et al., 2002, Vainionpää et al., 2006). These studies agree that dynamic loading is necessary to stimulate bone, but no agreement is achieved regarding the magnitude and frequency that such stimulation should have. Similar to transmissibility, it is not known if RMSa is significantly affected by osteoporosis on the lumbar spine.

Moreover, there is currently no information on how the lumbar curvature (lumbar lordosis) affects vibration transmissibility during physical activities. It has been suggested that vibration transmissibility is significantly affected during gait given a significant change on spinal curvatures (Bazrgari et al., 2008) and changes in the angle of lumbar extensor muscles due to ageing (Singh et al., 2011). However, it is not known if lumbar lordosis has a significant effect on vibration transmissibility during physical activity.

It was hypothesized that vibration transmission through the lumbar spine is significantly affected by osteoporosis and ageing during different types of physical activities. Further, it was hypothesized that lumbar lordosis is a significant determinant of the percentage of vibration transmitted through the spine.

4.2. Methods

4.2.1. Volunteers recruitment

Ethical approval was granted by the ethics committee of University of Roehampton (Appendix A). Male and female adults (healthy and with osteoporosis) were asked to volunteer from the general population in contact with University of Roehampton, community centres, Senior Citizen Clubs, markets, churches and libraries in the London area through a poster advertisement. Every volunteer was provided with an information sheet (Appendix A) after making the first contact with the principal investigator. All volunteers received an oral explanation of the study covering all contents in the information sheet and time was given for questions as recommended by the Declaration of

Helsinki (World Medical Association, 2008) section B paragraph 24. They were also assessed through an interview to determine their eligibility for the study. During the interview, a check list (Appendix A) was filled in by the principal investigator to verify the exclusion criteria. The volunteer's right to withdraw at any time during the study was explained before their written consent was obtained. All subjects were offered a copy of the consent form (Appendix A) they signed. At the end of the measurements volunteers were offered a debriefing form (Appendix A) with the identification number assigned to their data set (for data collection and storage purposes) ensuring anonymity.

4.2.2. Subjects

The appropriate sample size was determined by statistical power analysis, two tailed t test (Faul et al., 2007). This calculation was limited to previous research data, relying on healthy subjects and small sample sizes. An average value of previous published vibration transmissibility through the spine (2.71 ± 0.54) was used to estimate a clinically important difference (20%) due to osteoporosis (3.25 ± 0.54). Considering a significance level of 0.05 (α), an 80% test power with a standardised difference of 1 for the vibration transmissibility 17 was required. A higher sample size was recruited so that it would possible to detect any statistically significant changes in all variables. A total of 100 subjects were recruited, their characteristics can be seen in Table 4.2-1. Body mass was not significantly different between groups. The height was significantly different between the YH and OO groups but not between the OH and OO groups. The BMI was significantly different between the YH and OO groups only (Table 4.2-1).

Exclusion criteria consisted of having severe back or leg pain in the last 12 months that required medical treatment, severe rheumatological disorders, present spinal infections, previous or current dislocations or surgery of the spine and lower limbs. Volunteers were also excluded if they had been clinically diagnosed as obese, if they had any known history of previous osteoporotic fractures and if they were pregnant or allergic to ultrasound gel and adhesive tape. Volunteers were asked not to participate if they had an orthopaedic implant (a medical device that replaces part or a whole joint) or an electrically powered medical implant (for example a pacemaker, an implantable

defibrillator, a cochlear implant, neurostimulators or an insertable cardiac monitor). Any medical condition known during the interview which might interfere with normal function of the locomotor system, in the opinion of the investigator, was also a reason for exclusion. Maximum and minimum body mass indexes (BMI) were also a reason to enable subjects to participate in the study. The BMI restrictions were based on the World Health Organization BMI classification regarding underweight and obese limits. Subjects with a BMI below 18.5 kg/m^{2}or above 29.99 kg/m^2 were excluded. Given that the peak bone mass is reached at approximately 30 years old and that bone mass consistently decreases in both women and men after 55 years old (Figure 2.1-2), subjects were included in the research study if they were between 25 and 35 years old or older than 55. This ensured the inclusion of samples of the population with peak bone mass as well as older people with clear reduced bone mass due to age or osteoporosis.

Table 4.2-1 Subjects characteristics, mean (SD)

Characteristics	Groups according to BMD		
	YH	OH	OO
Age	29 (3.5)	65(8.1)	67(7.5)
Mass (kg)	69.65(11.57)	68.01(8.6)	65.16(8.42)
Height (m)	1.71(0.10)	1.65(0.07)	1.61(0.06)
BMI (kg/m^2)	23.53(2.50)	24.70(2.65)	23.04(2.67)
T-score	0.04(0.71)	-0.35(0.48)	-1.76(0.79)
Number of subjects	34	23	43
Female	16	19	41
Male	18	4	2

Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

4.2.3. Experimental conditions

Measurements took place at Whitelands College, University of Roehampton. Facilities that were used include the Biomechanics laboratory, a corridor outside it (33 m in length and 2 m wide) and a staircase consisting of 15 steps of normal height and 1.19 m wide, with a continuous hand rail on both sides (Figure 3.2-2).

Subjects were asked to wear loose clothing and the shoes that they used most of the time excluding high heels and sandals. The time taken to prepare and evaluate each volunteer was approximately

150 minutes. The stairs and corridor were temporarily closed to public use while in use. Also a disposable gown was provided to each participant to wear during the sensor attachment process. The gown enabled access to the sensors without exposing the volunteer's body. Once sensor attachment was concluded, the volunteer was able to wear the upper garments. Temperature of the Biomechanics laboratory was kept at the volunteer's preference to allow comfortable measurements.

A Broad Ultrasound Attenuation (BUA) test was performed to determine the T-score of each subject. This was used to identify those subjects with normal, osteopenic or osteoporotic skeleton. An ultrasound scanner (M-turbo®, FUJIFILM Sonosite Inc.) was used to allow the verification of spinous processes while subjects were lying down on a clinical examination bed facing downwards. An electromagnetic tracking device (3SPACE FASTRAK®, Polhemus Inc.) was used with its dedicated Motion Traking System software (Lee, 2005) through a desktop computer to record seven locations over the spine and the participant's spine curvature in a three dimensional space. Wireless inertial sensors (Wireless InertiaCube3™, Intersense Inc.) consisting of accelerometers and gyroscopes in three axes were used to measure acceleration and sensor inclination to the vertical through their dedicated IsPlot software (D'Anuono, 2010) and a desktop computer which was put on a trolley to enable its transportation through the testing areas. All walking paths had two wireless time gates (Smartspeed™, Fusion Sport Propietary Ltd.) located at known distances in order to calculate the participant's time taken to complete each trial. These gates were controlled through their dedicated personal digital assistant device (PDA) and software (Fusion Sport Propietary Ltd., 2010). The recorded times were used to calculate average walking speed. A mechanical weighing scale (CMS Weighing Equipment Ltd.) was used to measure each participant's mass in kilograms and a stadiometer (The Leicester height measure, Seca Ltd.) was used to measure their height in meters. Data processing and analysis was done through Excel® (Microsoft Corporation, 2007) and Matlab® version R2010b (The Mathworks Inc., 2010). Statistics were determined through IBM®SPSS® statistical software version 17.0 (SPSS Inc., 2009) and Matlab®.

4.2.4. Measurements

All subjects underwent BUA test to determine their BMD. Firstly this test was done one time on left and right ankle, once the non-dominant leg (ankle with the lowest BUA index) was found the test was done two more times on it. An average T-score was calculated for each subject. World Health Organization guidelines (WHO, 1994) were used to divide the groups into three according to their average T-score on the non-dominant ankle: normal density, osteopenia and osteoporosis.

In order to determine the spine curvature a local spine coordinate system of each subject had to be determined. For this, seven points of the spine were found through palpation and marked with a water marker to be employed in subsequent digitisations. The ultrasound scanner was employed to verify the location of spinous processes. This process required an examination of at least 30 minutes as the location of every spinous process was verified first through palpation and then via the ultrasound scanner. The Fastrak® electromagnetic tracking device was used to digitise these seven points of the body: first thoracic vertebral spinous process (T1), eighth thoracic vertebral spinous process (T8), twelfth thoracic vertebral spinous process (T12), first lumbar vertebral spinous process (L1), fifth lumbar vertebral spinous process (L5), and right and left posterior superior iliac spine (RPSIS and LPSIS respectively) (Figure 4.2-1).

In order to measure height, subjects were asked to stand barefoot with their heels together and touching the backstop of the stadiometer's base. Similarly, they were asked to have their legs straight with relaxed shoulders and arms on their sides. Their heads were positioned gently in the Frankfurt plane (imaginary horizontal line from the ear hole to the lower border or the eye). Then the height was read in meters and up to the last completed millimetre.

In order to measure body mass, the poise bar of the scale was secured, subjects were asked to stand on the scale barefoot in a relaxed position with their arms on their sides and remain still while the poised bar was released and moved until reaching mechanical balance. Body mass was read in kilograms and up to the last completed 100 grams.

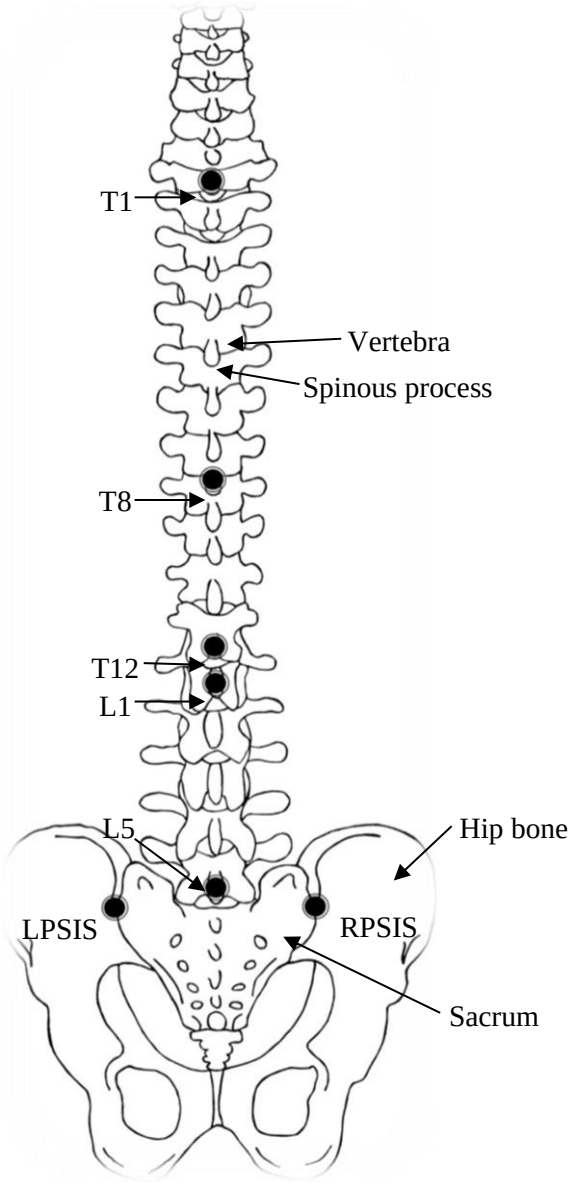

Figure 4.2-1 Seven points digitized to determine a local spine coordinate system and spine curvature. First thoracic vertebra (T1), eighth thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1), left posterior superior iliac spine (LPSIS), right posterior superior iliac spine (RPSIS)

Subjects were asked to stand in a relaxed usual posture. The seven points were marked with a tip attached to the electromagnetic sensor. This digitisation process was made three times to calculate the mean. The positions of T1, RPSIS and LPSIS were employed to establish the local coordinate system of the spine. All other points were used to calculate the spine curvature (Singh et al., 2010).

Vibration transmitted through the spine was examined using four InertiaCube3™ inertial sensors. One sensor over each of the following locations: T1, T8, T12 and S1 (Figure 4.2-2). Only data from the sensors over T12 and S1 was used for this part of the study. It was possible to calculate transmitted vibration when considering one sensor located over an output vertebra and another sensor located over an input vertebra, here T12 acted as output while S1 as input. All sensors were aligned with the long axis of the spine and attached to the subject's skin with double sided adhesive tape. The accelerometers detected linear acceleration and the gyroscopes sensed angular rotation, simultaneously.

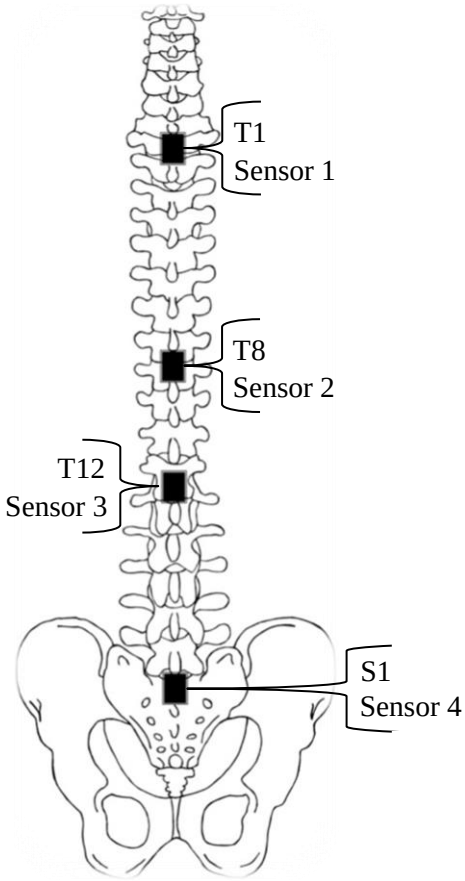

Figure 4.2-2 Location of inertial sensors over the spine. First thoracic vertebra (T1), eighth thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1)

A nudge test to correct for the skin-sensor interface movement was performed four times for each sensor. First, subjects were asked to ascend (a) and descend (d) stairs. Secondly, subjects

performed walking along a straight line (w) and lastly through a path consisting of combined turning and walking (m). The combined path consisted of straight walking separated by four right turns and four left turns and was considered to be completed until 60 steps were recorded in total (Figure 4.2-3). All physical activities were performed at self-selected normal walking speed (NWS) and repeated three times. Timing gates located at each path were used to measure the time that the subject took to complete the walking trial. A rest was given between trials to prevent fatigue.

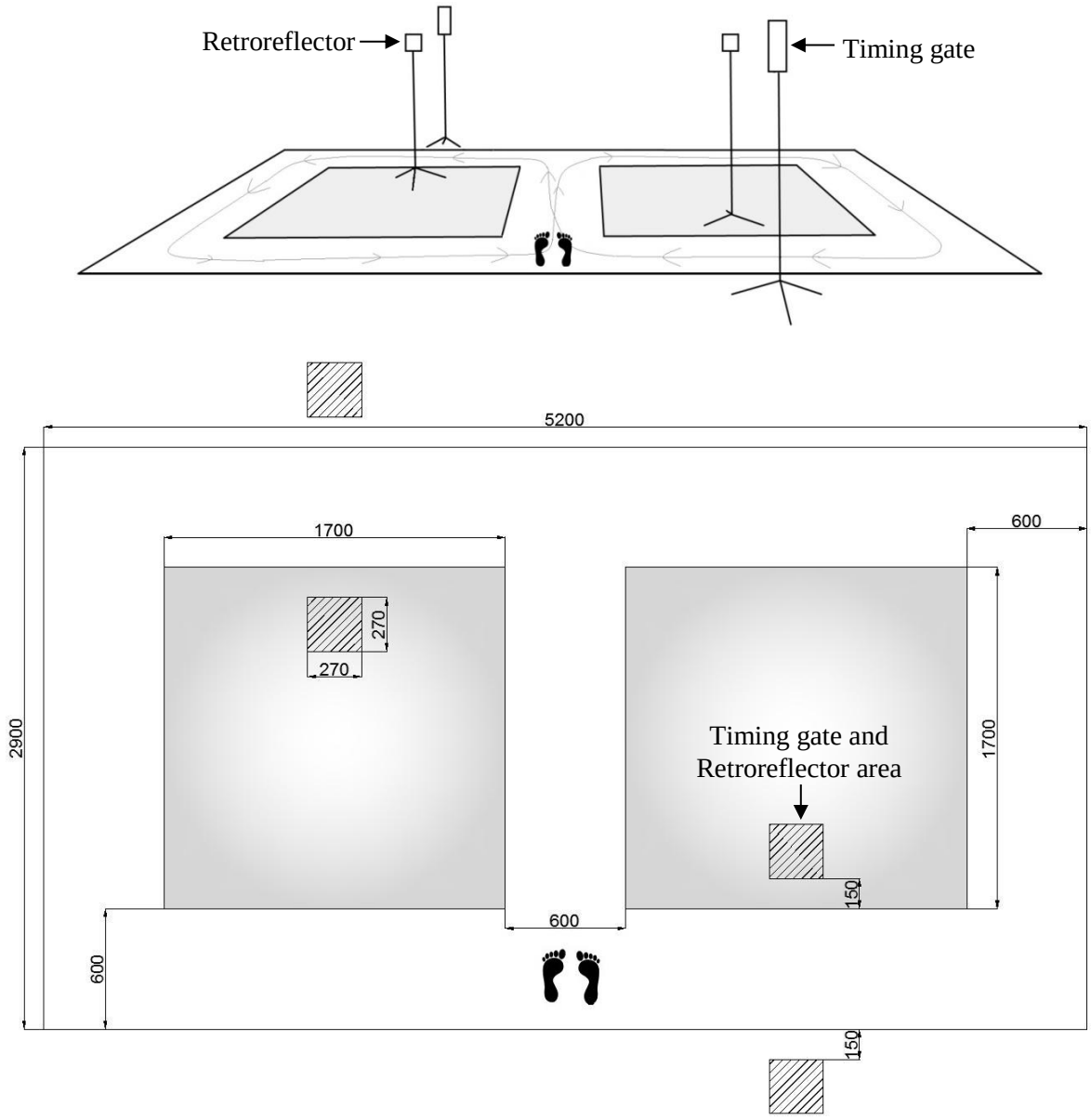

Figure 4.2-3 Combined walking and turning path with time gates on place and subject feet in the start position. Dimensions in mm

4.2.5. Data processing and analysis

Data was analysed using custom made scripts in Matlab®. Raw sensor acceleration and angular rate signals (sampled at 110 Hz) were low pass filtered at 20 Hz with a zero phase 5th order Butterworth algorithm. A maximum of 20 Hz was selected because human gait frequency content was previously reported to lie between 0.75 and 9.8 Hz (Angeloni et al., 1994, Cappozzo, 1982). This study explored transmissibility of vibration up to 8 Hz due to the natural frequency of the skin-sensor interface obtained through the model (Kitazaki and Griffin, 1995). Transmissibility of vertical vibration along the spine was estimated as the ratio of the power spectral density (PSD) of the output (T12) over the PSD of the input (S1) and over the frequency range of 0.5 to 8 Hz (Figure 4.2-4). Transmissibility was calculated for all physical activities and for each group (according to their bone health). Three different measures were used to describe the nature of the vibration transmitted through the lumbar spine: mean maximum transmissibility at frequency intervals ($0.5 \leq f \leq 2$, $2 < f \leq 4$, $4 < f \leq 6$ and $6 < f \leq 8$ Hz), mean maximum transmissibility at maximum acceleration PSD (maxT@maxPSD) and RMSa. Mean maximum transmissibility with a 95% confidence interval was determined for each frequency interval and calculated for each subject. Intervals helped to identify specific frequencies at which the highest transmissibility of vibration was obtained. Helliwell (1989) suggested the use of a single transmissibility value at maximum acceleration PSD in order to assess the amplification and attenuation properties of the spinal column with a single value. Therefore mean maxT@maxPSD was calculated for each subject and for all physical activities to evaluate the feasibility of employing a single value of transmissibility to express the dynamic response of the spine during physical activity. In order to measure the magnitude of transmitted vibration, acceleration spectral density ($(m/s^2)^2/Hz$) and RMSa were calculated. These calculations were made with vertical acceleration corrected for skin movement and sensor inclination at T12 and S1 for each trial.

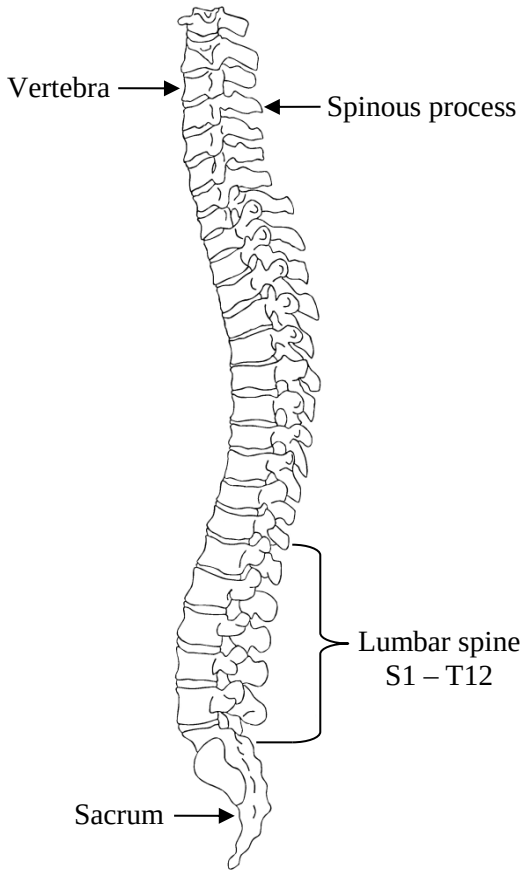

Figure 4.2-4 Transmissibility case based on normal lumbar spine curvature. Twelfth thoracic vertebra (T12), first sacral vertebra (S1)

4.2.6. Statistical analysis

Statistics were determined with IBM®SPSS® statistical software. All data was significantly non normal as determined by Kolmogorov-Smirnov tests and visual exploration through histograms against a normal curve, for each group and each physical activity. To test the hypotheses that mean maximum transmissibility at frequency intervals, maxT@maxPSD and RMSa are significantly different between physical activities and between groups, Kruskal-Wallis ANOVA was employed with a 0.05 significance level. Post hoc tests were performed with a Bonferroni correction. In order to determine the variables that are significant contributors of transmissibility, a forward multiple regression analysis was performed. Models to predict mean maxT@maxPSD were determined for each group. Mean maxT@maxPSD was chosen as it provides a single transmissibility value.

Lumbar lordosis, RMSa, age, walking speed (WS), BMI, T-score and gender were selected as predictors.

4.3. Results

4.3.1. Walking speed, skin correction factors and spine curvature

Mean walking speeds for each physical activity are presented in Table 4.3-1. Skin-sensor interface movement correction was done individually and for each sensor. Mean damping factors and natural frequencies utilized to correct for the skin-sensor interfaces movement are presented in Table 4.3-2. Not all subjects had the same quantity of sensors during trials due to technical problems or because the subject did not perform a particular trial, details of which can be seen in Appendix B. The lowest mean natural frequency was found for the skin-sensor interface of S1. Lumbar lordosis was not significantly different between groups. Mean spine curvatures are presented in Figure 4.3-1.

Table 4.3-1 Self selected walking speeds for all physical activities and groups, mean (SD)

	Walking speeds (m/s) for different physical activities			
Group	**Straight walking (w)**	**Walking and turning (m)**	**Ascending stairs (a)**	**Descending stairs (d)**
YH	1.659 (0.199)	1.337 (0.198)	0.584 (0.077)	0.688 (0.107)
OO	1.625 (0.264)	1.192 (0.196)	0.591 (0.144)	0.668 (0.179)
OH	1.752 (0.216)	1.325 (0.226)	0.590 (0.060)	0.697 (0.148)

Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

Table 4.3-2 Correction fators for skin-sensor interfaces over thespine, all groups, mean (SD)

Group	**Location**	**T12**	**S1**
YH	***fn* (Hz)**	16.452 (2.711)	12.615 (2.33)
	ζ	0.207 (0.127)	0.186 (0.079)
OH	***fn* (Hz)**	17.539 (3.898)	12.554 (2.320)
	ζ	0.405 (0.058)	0.325 (0.072)
OO	***fn* (Hz)**	15.853 (2.871)	13.150 (4.978)
	ζ	0.366 (0.072)	0.341 (0.062)

Natural frequency (f_n), damping ratio (ζ), young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), first sacral vertebra (S1)

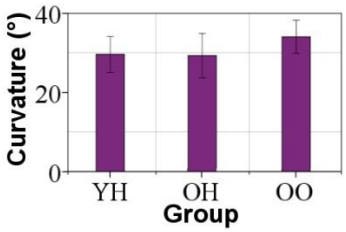

Figure 4.3-1 Lumbar lordosis between groups. Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

4.3.2. RMS acceleration

Significant differences in RMSa were found between groups with different BMD (Figure 4.3-2). Similarly, significant differences in RMSa were found between different types of physical activities. The magnitude of RMSa was significantly different between *w* and *m*, *a* and *d*, *w* and *a*, and between *m* and *d*, at both locations and groups. In addition, *w* and *d* produced significantly different RMSa magnitudes at both spine levels for the YH spine and at T12 only for the OO spine. Both levels of the OO spine received significantly different RMSa magnitudes during *m* and a. In contrast, this was an effect that was not observed for the healthy spines. Significant differences in RMSa between groups according to their BMD were also identified, these are indicated by the name of the groups followed by an asterisk (Figure 4.3-2). The magnitude of RMSa was the same between the YH and OO spines during *a* at S1 and during *w* and *a* at T12. The older spines received a significantly different magnitude of vibration during *m* at T12 and during *a* at S1 only. The OH spine received the same magnitude of vibration as the YH spine at T12 and S1 during *w* and *m*.

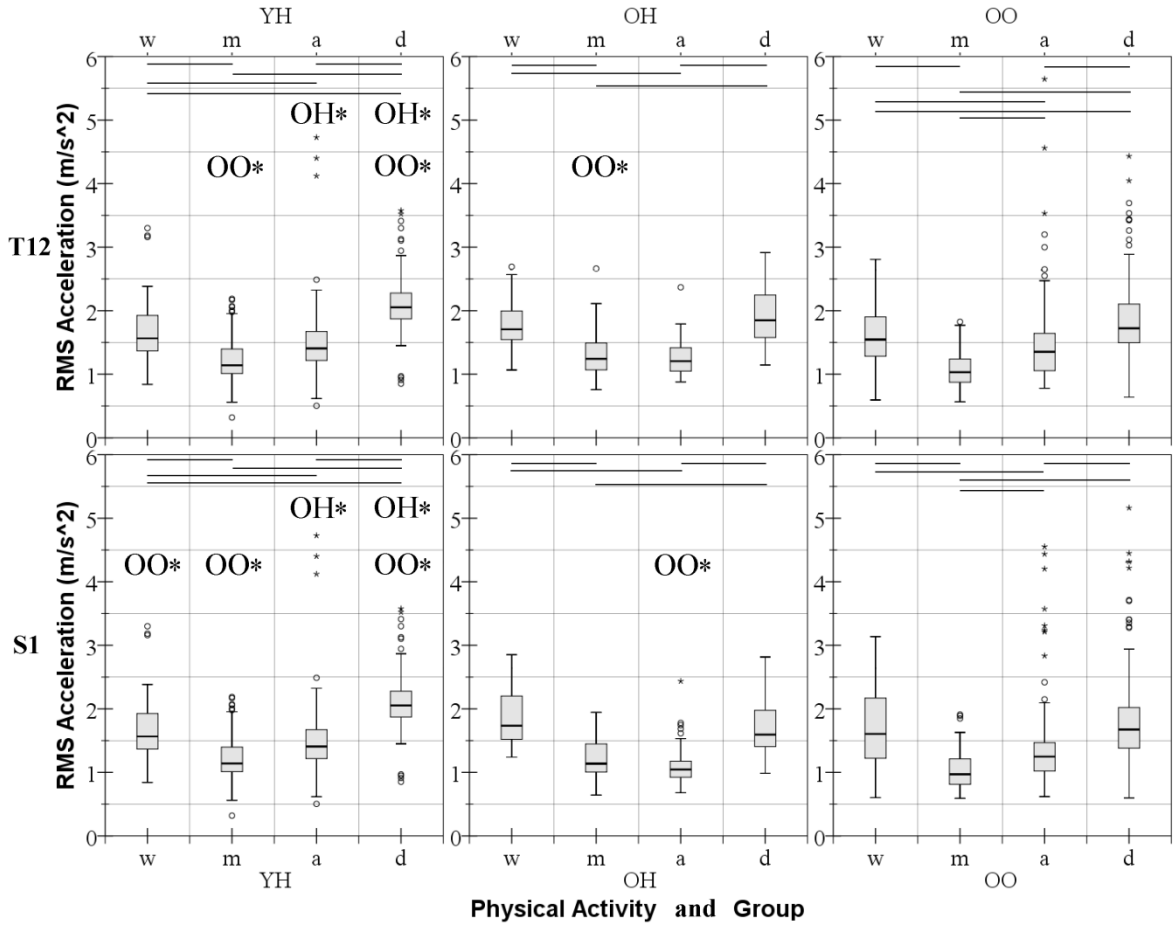

Figure 4.3-2 Root mean square (RMS) acceleration at the twelfth thoracic vertebra (T12) and first sacral vertebra (S1). Comparison between physical activities and groups.─ or * = significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

4.3.3. Transmissibility overview

Mean transmissibility and acceleration PSD $((m/s^2)^2/Hz)$ are presented with 95% confidence intervals for straight walking at NWS for comparison of the three groups according to their BMD (Figure 4.3-3). The transmissibility curves show that at some frequency intervals there is amplification and at other attenuation. The shapes of the transmissibility curves at first seem to be the same. When considering the limit of 100% transmissibility the significance of a curve being near, below or above this limit suggests that osteoporosis and ageing may have a significant influence over transmissibility at specific frequencies.

Mean acceleration PSD clearly shows that the magnitude of the vibration transmitted from S1 to T12 is amplified between 1.5 and 3 Hz during all physical activities regardless of ageing and osteoporosis and during all physical activities. At all other frequencies the PSD method detected a very small magnitude of acceleration. These transmissibility curves were summarized through the maximum transmissibility at frequency intervals. The amplitude of the vibration transmitted was summarized through RMSa.

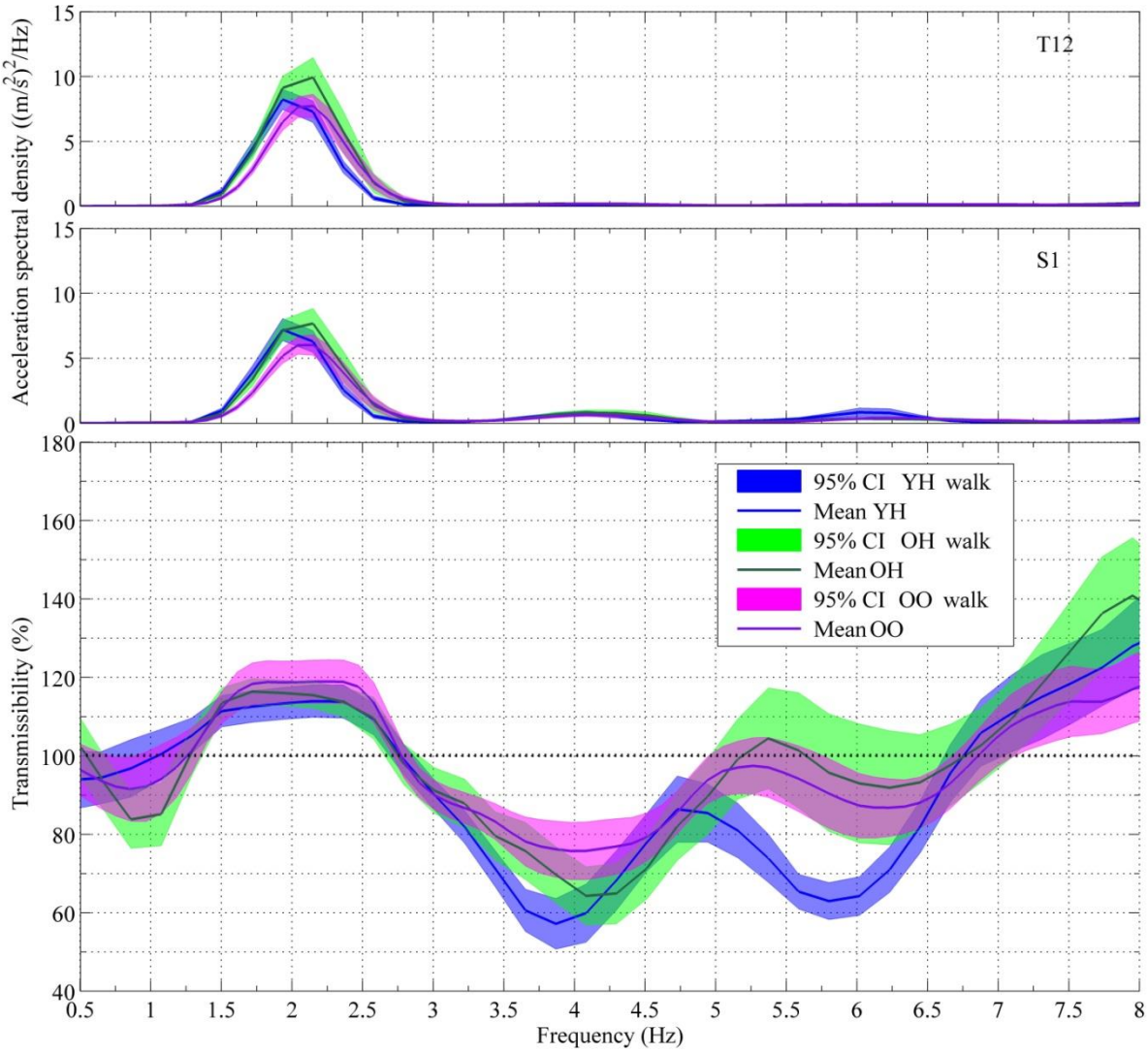

Figure 4.3-3 Input (S1) and output (T12) acceleration PSD and transmissibility, walking in a straight line during normal walking speed for all groups. First sacral vertebra (S1), twelfth thoracic vertebra (T12), confidence interval (CI)

Mean transmissibility and acceleration PSD ($(m/s^2)^2$/Hz) are presented with 95% confidence intervals for comparing physical activities at NWS for the YH group (Figure 4.3-4). Vibration transmitted by the lumbar spine during *w* was amplified above approximately 1.75 Hz to be later attenuated after approximately 2.75 Hz. Combined walking and turning had similar changes in amplification and attenuation through the frequency interval studied. Stair ascent and descent present fewer transmissibility oscillations yet frequency intervals at which these reached similar magnitudes as *w* and *m*. This example for the YH lumbar spine (Figure 4.3-4) shows that different physical activities produce different transmissibility at different frequency intervals.

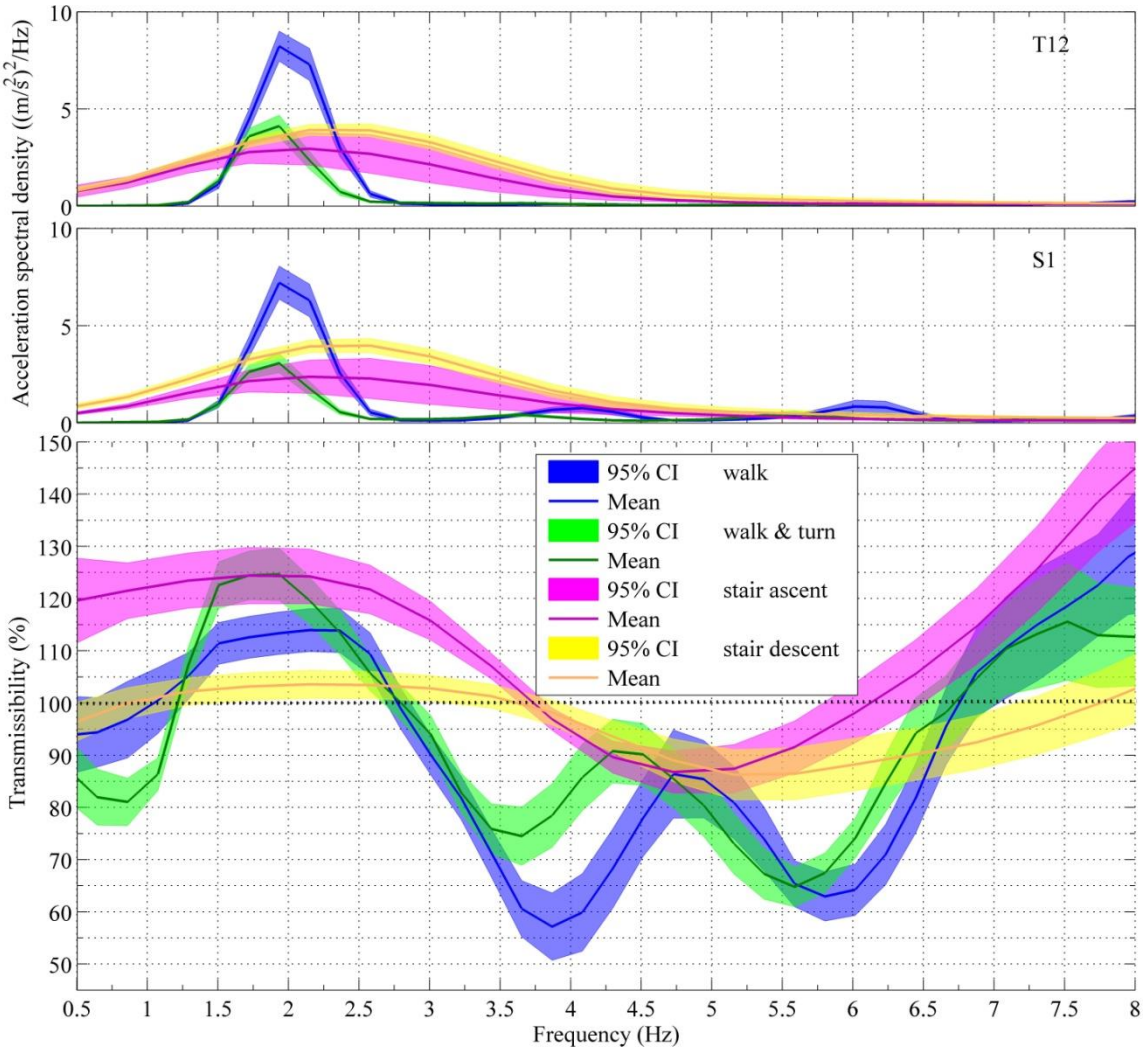

Figure 4.3-4 Input (S1) and output (T12) acceleration PSD and transmissibility duringnormal walking speed (NWS) at all physical activities for the young and healthy group. Confidence interval (CI), first sacral vertebra (S1), twelfth thoracic vertebra (T12)

4.3.4. Mean maximum transmissibility at frequency intervals

Significant differences in mean maximum transmissibility between different physical activities were found (Figure 4.3-5). Significant differences in transmissibility were seen between groups of different age and BMD (Figure 4.3-5). Mean maximum transmissibility with 95% confidence intervals were determined for 4 different frequency intervals: $0.5 \leq f \leq 2$, $2 < f \leq 4$, $4 < f \leq 6$ and $6 < f \leq 8$ Hz. Overall, the lumbar spine generally amplified vibration. Mean maximum transmissibility for the YH spine was significantly different between *m* and *d* at all frequency intervals. Mean maximum transmissibility was not significantly different between *w* and *m* at all frequencies for the YH spine. Mean maximum transmissibility was found to be not significantly different between *w* and *m* at the intervals $0.5 \leq f \leq 2$ and $6 < f \leq 8$ for the older spines. During *a* and *d* the vibration transmitted by the older spines was not significantly different at all frequency intervals. Mean maximum transmissibility was significantly different between *m* and *d* at all frequency intervals for the OO spine while for the OH this was observed only between 0.5 and 4 Hz.

Mean maximum transmissibility was significantly different between groups at different frequency intervals (Figure 4.3-5). The YH spine transmitted significantly less vibration than the older spines at all frequency intervals during *d*. This behaviour was also observed during *m* ($2 < f \leq 6$ Hz) and during *a* ($4 < f \leq 6$ Hz). The OO spine transmitted significantly less vibration than the OH during *w* and *a* (from 6 to 8 Hz) and during *d* from 4 to 8 Hz. There were physical activities and frequency intervals at which ageing and osteoporosis had no significant effect on mean maximum transmissibility. All groups transmitted a similar percentage of vibration during *w* ($0.5 \leq f \leq 4$ Hz), *m* ($0.5 \leq f \leq 2$ Hz) and during *d* from 0.5 to 4 Hz (Figure 4.3-5). The YH lumbar spine amplified vertical vibration for all physical activities from 0.5 to 4 Hz. The YH spine had the greatest amplification (155%) during *w* at the $6 < f \leq 8$ frequency interval whereas the least transmissibility (95%) was observed during *d* at the $4 < f \leq 6$ frequency interval. The older lumbar spines amplified vertical vibration at all physical activities and for all frequencies studied. The greatest mean maximum transmissibility was observed always for the OH spine for all physical activities. The greatest mean maximum transmissibility at the highest frequency interval for the OH spine was

145% during *d* and 170% during w. The OO spine reached the greatest amplification (151%) during *m* at the 6< f ≤8 frequency interval whereas the least transmissibility (113%) was observed during *d* at the 4< f ≤6 frequency interval.

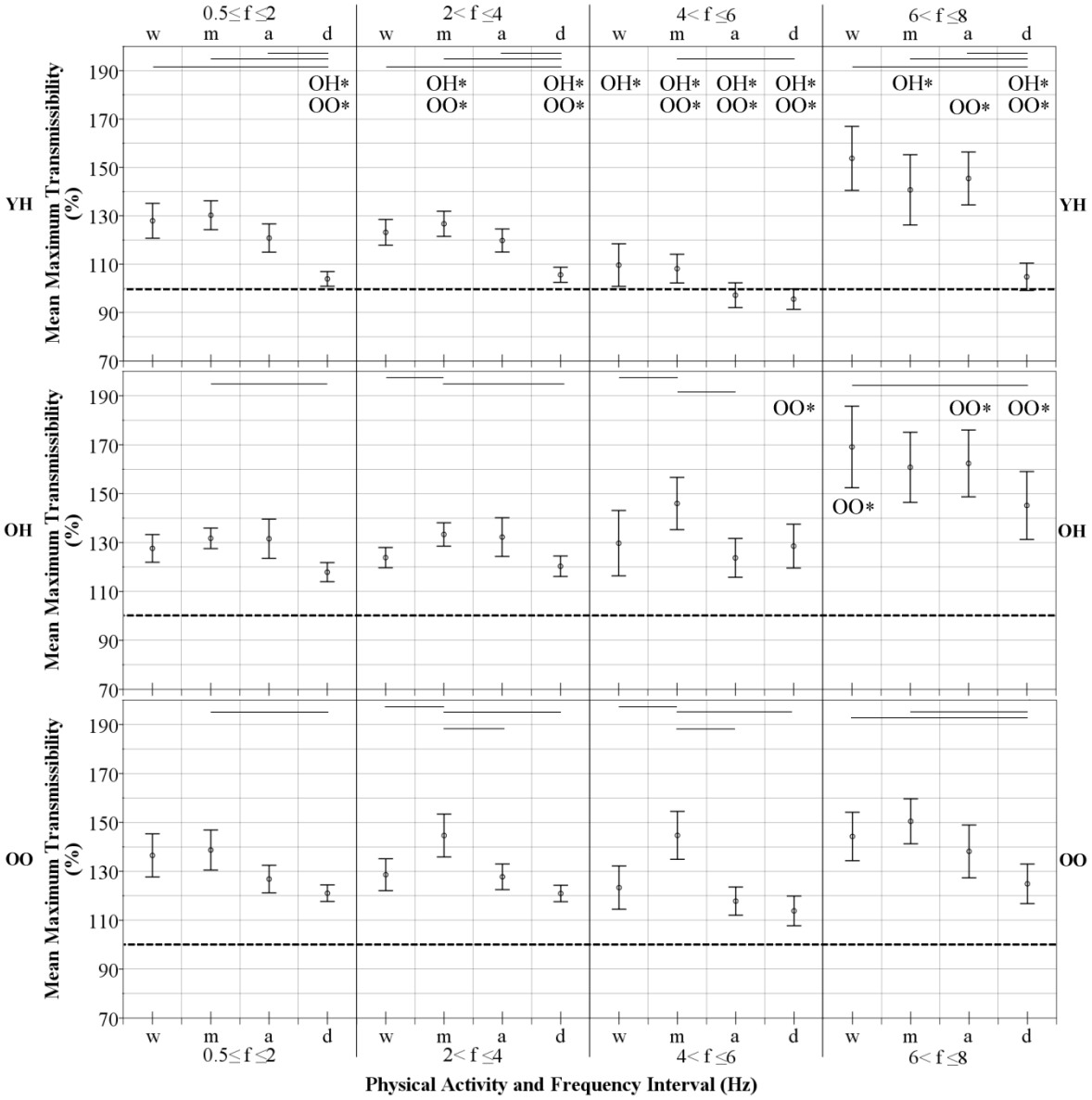

Figure 4.3-5 Mean maximum transmissibility from first sacral vertebra (S1) to twelfth thoracic vertebra (T12) at frequency intervals. ─ = significant difference. Dotted line= 100% transmissibility, attenuation below and amplification above it. Young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

4.3.5. Mean maximum transmissibility at maximum acceleration PSD

Mean maximum transmissibility at maximum acceleration PSD was significantly different between different physical activities at YH an OH spines (Figure 4.3-6). The lumbar spine also indicated amplification of vibration through maxT@maxPSD. The YH spine transmitted a significantly different percentage of vibration at maximum acceleration PSD during *a* and *d*. During *w* and *m* the OH spine produced significantly different maxT@maxPSD. All physical activities produced the same maxT@maxPSD for the OO spine. The lowest maxT@maxPSD for the YH lumbar spine was observed during *d* (103%). Conversely, the highest amplification was observed during *a* and *m* (124%). The older spines amplified vibration during all physical activities. The lowest maxT@maxPSD for the older spines was observed during *w* and *d*, with a slightly higher magnitude for the OO spine (119 % in comparison with 117% for the OH spine). The highest amplification achieved by the OH spine was during *a* (130%) whereas this was observed during *m* for the OO spine (129%). The YH spine transmitted significantly less vibration than the OH and OO spines during *d* only. Significant differences between groups according to their BMD are indicated by the name of the groups followed by an asterisk (Figure 4.3-6). Significant differences between groups based on maximum transmissibility at maximum acceleration PSD were present during *d* only. The YH spine transmitted significantly less vibration than the OH and OO spines. Overall amplification of vibration is indicated by maxT@maxPSD for the lumbar spine.

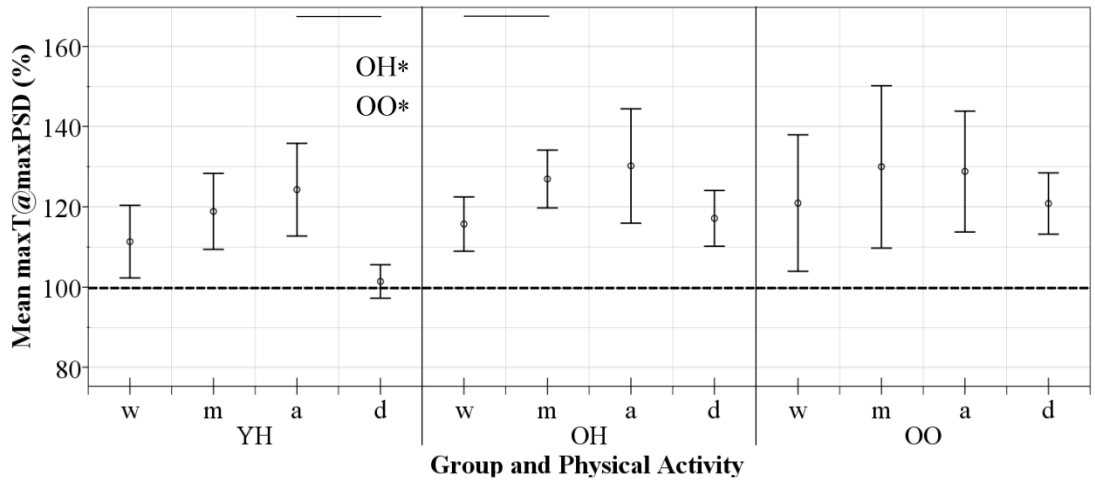

Figure 4.3-6 Mean maximum transmissibility at maximum acceleration PSD (maxT@maxPSD)during normal walking speed for the lumbar spine.─ or * = significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

4.3.6. Transmissibility predictors

Prediction of mean maximum transmissibility at maximum acceleration PSD dependant of RMSa, lumbar lordosis, BMD, age, gender and BMI was performed (Table 4.3-3). Gender was considered as a predictor only for the young and healthy spine due to the equivalent distribution of gender among this group only. Based on the value of the correlation coefficient, RMSa at S1 was the most important and significant predictor for the lumbar spine of all groups. 37.9% and 41.4% of lumbar spine's transmissibility variability of the YH and OH groups respectively was explained in terms of predictors used and included the lumbar lordosis. T-score and walking speed were important and significant only for the OH spine. Conversely only 15.3% of transmissibility variability was explained by the predictors for the OO lumbar spine and lumbar lordosis was not a predictor. BMI was significant and important only for the OO group. Age was an important predictor of transmissibility for all groups. The specific transmissibility variability explained by lumbar lordosis was 2.2% and 2.5% for the YH and OH groups respectively.

Table 4.3-3 Models to predict mean maximum transmissibility at maximum acceleration PSD for the lumbar spine during normal walking speed

Group	r^2	Residual	Variability explained	Model Term	Coefficient	Significance	Importance
	0.379	4.133	37.9%	Intercept	1.445	0.000	
				RMSa at S1	-0.141	0.000	0.278
YH				Female	0.175	0.000	0.267
			2.2%	Lumbar Lordosis	0.003	0.030	0.230
				Age	-0.007	0.147	0.225
	0.414	1.676	41.4%	Intercept	1.773	0.000	
				RMSa at S1	-0.432	0.000	0.204
				T-score	-0.121	0.000	0.168
OH				RMSa atT12	0.231	0.001	0.165
				Age	-0.005	0.011	0.158
			2.5%	Lumbar Lordosis	-0.002	0.050	0.153
				WS	0.006	0.070	0.152
	0.153	11.085	15.3%	Intercept	1.190	0.000	
				RMSa at S1	-0.237	0.000	0.255
OO				BMI	0.028	0.000	0.254
				Age	-0.008	0.002	0.248
				RMSa at T12	0.167	0.021	0.243

Root mean square acceleration (RMSa), young and healthy (YH), older healthy (OH), older osteoporotic (OO), body mass index (BMI), first sacral vertebra (S1), twelfth thoracic vertebra (T12), walking speed (WS)

4.4. Discussion

The main aim of this study was to assess the potential effect of osteoporosis on vibration transmission of the lumbar spine during physical activity. Similarly, the effect of ageing was also investigated. The magnitude and percentage of vibration transmitted by the lumbar spine are significantly affected by ageing. Osteoporosis has also an effect but is different to that of ageing. Physical activities produce vibration with multiple magnitude and frequency components. Lumbar lordosis has a small but important and significant association with the percentage of vibration transmitted through the spine but only for individuals without osteoporosis.

Mean maximum transmissibility at different frequency intervals provided evidence that the lumbar spine amplifies vibration during most physical activities tested. This amplification may help to stimulate bone growth and might explain why fractures are not common in this region of the spine. Transmissibility of healthy spines has been previously reported during walking from S1 to the second thoracic vertebra (T2) (Smeathers, 1989b) and from sacrum to T2 (Smeathers, 1989a) from 1 to 40 Hz. Comparison with these previous studies is difficult since they measured transmissibility in a single subject and in a section of the spine that included the lumbar and thoracic spines. However their results also suggested amplification below 8 Hz. The major contribution of this study is that spine vibration transmission during physical activity has been measured on individuals with osteoporosis for the first time.

Ageing increases vibration transmission at frequencies greater than 4 Hz (Figure 4.3-5). It is hypothesized that this amplification is due to the stronger muscle contraction required to maintain balance or produce motion during physical activities. Stronger muscle contraction indicates that greater loads are exerted on the spine (Izzo et al., 2013). Osteoporosis decreases stiffness at frequencies higher than 6 Hz (Figure 4.3-5). This attenuation acts against the amplification seen due to ageing. This attenuation may be due to thinning in the trabeculae of vertebrae and the consequent reduction in BMD as a consequence osteoporosis. As vibration is transmitted through the human body it is stored, dissipated and distorted (Collins and Whittle, 1989). Vibration components of different frequencies and magnitudes travel at different speeds through the body,

depending on the material properties of tissue. Consequently, the overall mechanical properties of the spine are determined by the size and shape of vertebrae and intervertebral discs and by the material properties of every tissue of the spine (cartilage, muscle, bone, tendon and ligaments). Decreased spinal damping in people with osteoporosis has been observed previously (Bhattacharya et al., 2010, Orkoula et al., 2012, Yerramshetty and Akkus, 2008). Yet, this effect may be more related to ageing. This may have not been observed before because most studies into osteoporosis are performed with individuals with osteopenia and osteoporosis, not with healthy older adults. Stiffer tissue has been suggested as secondary to the loss of collagen (Shuster, 2005, Castelo-Branco et al., 1994). This stiffer tissue may justify the increased vibration transmissibility amplification in the lumbar spine. Also, vibration transmissibility at the lumbar spine may be amplified by the natural larger size of lumbar vertebrae at the spine (Tortora and Grabowski, 2003). In summary, the combination of bone stiffness increment (reduced damping) and structural changes (size and geometry) due to ageing may explain the prevalence of vibration transmissibility amplification of the lumbar spine regardless of osteoporosis.

Ageing decreases the magnitude of the vibration (RMSa) transmitted during stair negotiation. However, during walking (straight and with turning) RMSa was not affected (Figure 4.3-2). It is not clear if walking may be a better choice of activity for inducing bone growth in contrast with stair negotiation. Osteoporosis has a negligible effect on RMSa when compared with ageing. This is not associated with walking speed since all subjects walked at a similar self-selected normal comfortable speed regardless of age and osteoporosis. Although the percentage of vibration transmitted is amplified, the magnitude of that vibration transmitted seems to be attenuated from sacrum up to T12 at specific frequencies (Figure 4.3-2). In general, walking (straight and with turning) as well as stair descent and ascent were pairs of physical activities that produced statistically similar RMSa magnitudes. All other paired comparisons of physical activities produced significantly different RMSa magnitudes. The importance of these results is that the measurement of RMSa is capable of providing information on the magnitude of vibration transmitted to different locations of the lumbar spine and during different physical activities. The measurement of RMSa

has also provided evidence that the older spine may be receiving vibration magnitude similar to a young and healthy spine at the levels of S1 and T12 and that this is dependent on physical activity. The relationship between RMSa dose and bone metabolism response cannot be determined from this study. However, previous studies support the hypothesis that bone is sensitive to mechanical stimulation produced during physical activity (Morgado Ramírez et al., 2013c, Johnell and Kanis, 2006). Dynamic loading from 0.5 to 2 Hz has been shown to have an effective osteogenic effect (Morgado Ramírez et al., 2013c). This study has provided evidence that physical activity produces mechanical stimulation of low frequency (below 8 Hz). However, it is not possible to compare RMSa reported here with previously reported magnitude of stimulation in terms of με, N and g forces.

Lumbar lordosis has a small but important and significant contribution to the prediction of transmissibility for the young and older healthy groups. It is suggested that transmissibility is greater with greater lumbar lordosis for young individuals while for older subjects this is the opposite. According to the model obtained, the older the individual the lower transmissibility is through the lumbar spine. T-score had a surprisingly negative effect on transmissibility of vibration for older healthy individuals (Table 4.3-3). It is very likely that the remaining transmissibility variability that was not possible to explain is related to geometric and material properties of the spine that are not possible to measure in vivo during physical activity. The fact that no more than 37.9% of transmissibility was predicted by the measurements performed suggests that an individual assessment of transmissibility might be more appropriate.

It is interesting to note that the lumbar spine produced the greatest amplification of vibration during straight walking for all groups. If we assume that the mean maximum transmissibility at frequency intervals achieved by the YH lumbar spine is a threshold to determine the effect of ageing and osteoporosis, significant differences were found when compared with the OH and OO spines. Frequency intervals at which this threshold was significantly exceeded were associated with ageing and osteoporosis. These results also suggest that there are frequency intervals and physical

activities at which the lumbar spine receives the same percentage of vibration regardless of ageing and osteoporosis (Figure 4.3-5). The response of the lumbar spine with ageing and osteoporosis is a challenge for clinicians wishing to find optimal physical activities to safely stimulate the older healthy and osteoporotic lumbar spine (Hamilton et al., 2010a). Current specialized medical imaging techniques are capable of determining volumetric bone strength measurements (Griffith and Genant, 2012, Cheung and Giangregorio, 2012). Mean maximum transmissibility could be correlated with bone strength determined through high resolution imaging. Once this correlation is determined and bone strength can be predicted solely by mean maximum transmissibility, this technique could be employed to objectively characterize and identify optimal physical activities to treat osteoporosis safely and effectively at the lumbar spine. Similarly, it is possible that the performance of physical activity has an accumulative outcome on the effect of bone. For example, it has previously been observed that walking from 6 to 12 months does not preserve BMD in the lumbar spine of postmenopausal women (Martyn-St James and Carroll, 2008). This may indicate that other forms of physical activity capable of maintaining or increasing BMD have to be identified. This could be tested in future research employing vibration transmission analysis. Finally, given the anatomical differences between lumbar and thoracic spines, further research is also necessary to determine the vibration transmission of the thoracic spine. The lumbar and thoracic spines may have significantly different mechanical responses during gait given that age related vertebral geometry and strength changes have been found to be greater at the lumbar than thoracic spine (Samelson et al., 2012). The following chapter studies vibration transmission of the thoracic spine.

Vibration transmitted through the spine has not previously been characterised through maxT@maxPSD. Helliwell (1989) suggested that limiting transmissibility to the maximum acceleration PSD may be sufficient for analysis. Here, it has been shown that some information on the effect of physical activity, ageing and osteoporosis is lost by not taking into account other frequencies in comparison with a single frequency at which maximum acceleration PSD is found. Yet similar to maximum transmissibility at frequency intervals, maxT@maxPSD indicates an

amplification tendency during all physical activities and groups (Figure 4.3-6). The use of maximum transmissibility at frequency intervals is recommended over maxT@maxPSD due the greater information content regarding the effect of physical activity, ageing and osteoporosis observed at various frequencies.

Bohannon and Williams Andrews (2011) performed a meta analysis of self-selected normal walking speeds reported for healthy individuals from 20 to 99 years old. Young subjects (20 to 39 years old) walked at 1.39 m/s while older subjects (50 to 99 years old) walked at 1.19 m/s. Boyer et al. (2012) reported normal walking speed for young and healthy subjects (28 ± 4.9 years old) as well as for older healthy adults (57 ± 4.5 and 71.2 ± 4.4), 1.36 m/s and 1.48 m/s respectively. These values are below the straight walking speeds reported (Table 4.3-1) while being similar for the walking and turning trials. OH subjects walked faster than YH. Walking speeds during stairs ascent and descent have been reported for healthy individuals (young and older) and lie below the ones reported in this study. Self-selected walking speeds during stair ascent for young and healthy individuals have been reported to be 0.51 m/s (Kretz et al., 2008), 0.49 ± 0.05 m/s (Protopapadaki et al., 2007) and 0.38 m/s (Reid et al., 2010). Self-selected walking speeds during stair descent for young and healthy individuals have been reported to be 0.56 ± 0.06 m/s (Protopapadaki et al., 2007) and 0.53 ± 0.08 m/s (Cluff and Robertson, 2011). Reid et al. (2010) reported a self-selected walking speed during stair ascent for healthy older adults (65.5 ± 5.2 years old) as 0.45 m/s, which is also below the one reported here. It has been suggested previously that adults who maintain an active life manage to maintain walking speeds as younger adults (Boyer et al., 2012). The sample of the population employed in this study may have preserved an active life thus the walking speeds reported are slightly above what has been seen in the literature.

4.4.1. Limitations

The vibration amplification seen at the lumbar spine is specific for a determined walking distance. In addition, it is not known if that amplification will persevere with longer times of physical activities (greater stimulation over time). The contribution of muscular contraction to the vibration

amplification is a hypothesis only, further research is needed to draw a conclusion on the contribution of muscular contraction in spinal vibration transmission.

The T-score was determined through QUS which provides only an approximate value for the general health of the skeleton. Dual X-ray absorptiometry is preferred for the clinical diagnosis of osteoporosis and drug treatment monitoring. The calculation of vibration transmissibility is an approximation due to the skin-sensor interface movement correction model (Kitazaki and Griffin, 1995, Morgado Ramírez et al., 2013c). Yet, this approximation is very close to what will be measured by inserting pins directly to bone (Kim et al., 1993, Ziegert and Lewis, 1978). Moreover, the skin-sensor interface movement correction requires individuals to be in a healthy weight to facilitate the attachment of inertial sensors to the spine. RMSa measurement provides the magnitude of vibration transmitted to locations of the spine at all frequencies and up to 20 Hz. The results of this study provide independent observations on the differences in vibration transmission due to osteoporosis, age and physical activity. It is not possible, from these results, to determine relationships between variables or whether a difference is driven by a third unmeasured variable. For example, smoking, alcohol consumption, level of physical activity throughout life, skinfold measurements, family history of fracture, pharmacological treatments and so on. Future research could help elucidate these relationships. Thus findings must be interpreted with caution. Large transmissibility variability was seen between subjects even when classified into groups according to their bone mineral density. Large transmissibility variability between healthy subjects has been reported previously (Kitazaki and Griffin, 1995). A greater sample of the population would be needed to account for variability between subjects. Other unmeasured variables which may affect vibration transmissibility are family history of osteoporosis, ethnicity, gender, alcohol and tobacco consumption habits as well as diet, risk of fracture, risk of falling, use of pharmacological treatments for osteoporosis and nonlinear analysis of gait such as dynamic stability and complexity. It was not possible to use gender as a predictor during the multiple regression analysis for the older spines (healthy and osteoporotic) since more female subjects participated in this study than male subjects. Similarly, alcohol consumption and smoking habits could not be considered since most

volunteers neither consumed alcohol nor smoked. Lastly, the physical activities tested are only a limited sample of everyday physical activities. Daily monitoring with inertial sensors may overcome this constraint.

4.5. Conclusion

Daily life physical activities produce vibration that is amplified by the lumbar spine at specific frequencies. This may help maintain the mechanical stimuli required for bone health, and explain the low incidence of vertebral fractures in the lumbar spine region. Ageing and osteoporosis affect the vibration transmission of the spine in different ways. Osteoporosis decreases vibration amplification during activities such as ascending and descending stairs at specific frequencies. The magnitude of acceleration experienced by the lumbar spine during walking is the least affected by ageing and osteoporosis. It is unclear if walking may be more effective in maintaining bone health compared to other activities examined in this study. Future research should examine the optimal dose of mechanical stimulus (as determined by the magnitude, frequency and percentage transmission of such vibration) required for stimulating bone growth.

4.6. Key findings

1. The lumbar spine amplifies vibration transmitted during physical activities at a self-selected normal walking speed.
2. Ageing increases vibration transmission at frequencies greater than 4 Hz.
3. Osteoporosis decreases stiffness (diminishes the amplification effect due to ageing) at frequencies higher than 6 Hz.
4. Ageing decreases the magnitude of the vibration transmitted during stair negotiation but not during straight walking and a combination of straight walking and turning.
5. Osteoporosis has a negligible effect on the magnitude of vibration measured when compared with ageing.
6. A percentage of vibration transmission of the lumbar spine is determined by its curvature during gait (below 2.5%).

CHAPTER 5 Vibration Transmission through the Thoracic Spine

5.1. Introduction

Vertebral fractures due to osteoporosis have a significant impact on daily life activities since they cause back pain, loss of height, deformity, immobility and reduced pulmonary function (IOF, 2012). Common vertebral fractures have been reported to occur most often at the thoracic spine (T5-T9, T7-T8) and at the junction of thoracic and lumbar spines (L5-T12) (Ravishankar, 2009, Waterloo et al., 2012). It has been suggested that vertebral fractures may occur during physical activity without serious symptoms, even when individuals have a BMD not classified as osteoporosis (Lems, 2007, Kanis et al., 2008). In contrast, there are studies that have found no significant effect of physical activity on vertebral fractures but rather a reduction of vertebral fracture risk (Moayyeri, 2008). Thus, the effect of physical activity on the incidence of vertebral fractures is unclear. Physical activities that have been identified to improve spinal BMD are sparse. These activities include walking, strength training and a combination of other physical activities (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Cheung and Giangregorio, 2012). Yet, BMD improvements through physical activity are site specific and induce less than 2% spine BMD increment (Hamilton et al., 2010a, Gómez-Cabello et al., 2012). A lack of understanding of the mechanical stimulus transmitted through the thoracic spine limits clinical approaches to manage osteoporosis through physical activity. We do not know the precise effects of ageing and osteoporosis on the signal transmission.

Heel strikes produce vibration that is transmitted through the body during gait (Collins and Whittle, 1989, Cappozzo, 1982). Bone responds to mechanical stimulation in the form of vibration (Skerry, 2008, Chen et al., 2010). The way this vibration is transmitted through the bone depends on its material and structural properties (Keller et al., 2000, Bediz et al., 2010, Bhattacharya et al., 2010, Kawchuk et al., 2009). This vibration can be measured in terms of the percentage transmitted

through the spine. Transmissibility is the ratio of the vibration measured between two points, is a function of frequency and has no units (Mansfield, 2005a). Transmissibility greater than 100% equals amplification and lower than 100% equals attenuation (Mansfield, 2005a). Transmissibility has been previously calculated at the whole spine of ten young and healthy subjects during walking, turning and stair negotiation (Morgado Ramírez et al., 2013c). Transmissibility is measured through inertial sensors attached to the spine with double sided adhesive tape (Saha and Lakes, 1977, Ziegert and Lewis, 1979, Hinz et al., 1988, Smeathers, 1989a, Smeathers, 1989b, Kim et al., 1993, Kitazaki and Griffin, 1995, Pankoke et al., 2001). Transmissibility may help understand the effect of physical activity at levels of the thoracic spine where vertebral fractures are common. Vibration transmissibility of the thoracic spine during physical activity has not been measured before in individuals with osteoporosis. The magnitude of vibration transmitted is calculated in terms of root mean square acceleration (RMSa) (Mansfield, 2005a, Asselin et al., 2011). Most studies have used animal models and a few humans to measure the magnitude of stimulation delivered to bone in terms of micro strains (με), Newtons (N) and acceleration (g forces and m/s^2) (Turner et al., 1994, Asselin et al., 2011, Al Nazer et al., 2012, Burr et al., 2002, Vainionpää et al., 2006). The effect of osteoporosis on RMSa of the thoracic spine during physical activity has not been reported.

There is currently no information on how the thoracic curvature (thoracic kyphosis) will affect vibration transmissibility during physical activities. It has been suggested that thoracic kyphosis has a strong linear relationship with spinal load profiles (Briggs et al., 2007, Morosano et al., 2011). Therefore vibration transmission may be significantly affected during gait given a significant change on thoracic curvature.

The lumbar and thoracic spines may have significantly different mechanical responses during gait given that age related vertebral geometry and strength changes have been found to be greater at the lumbar than thoracic spine (Samelson et al., 2012). Differences in vibration transmissibility between lumbar and thoracic spines have not been established.

It was hypothesized that (1) the magnitude of vibration transmitted is significantly affected, in locations where vertebral fractures are common, when comparing individuals with different BMD, (2) that there are significant differences in the percentage of vibration transmitted between lumbar and thoracic spines and (3) that thoracic kyphosis has an important and significant contribution to the transmission of vibration during physical activity.

5.2. Methods

Participants and measurements are the same as the previous chapter except for the location of the sensors and the section of the spine studied. Participants were divided into three groups: young and healthy (YH), older healthy (OH) and older with osteopenia and osteoporosis (OO) (Table 4.2-1).

Thoracic kyphosis was recorded through an electromagnetic tracking device (Singh et al., 2010). Three inertial sensors were attached with adhesive tape over the spinous process of the twelfth (T12), eighth (T8) and first (T1) thoracic vertebra (Figure 4.2-2). First, participants were asked to ascend (a) and descend (d) stairs consisting of 15 steps of normal height and 1.19 m wide with a continuous hand rail on both sides (Figure 3.2-2). Secondly, to perform walking along a straight line (w) and lastly through a path consisting of combined turning and walking (m) (Figure 4.2-3). Vertical acceleration and dynamic sensor inclination were collected throughout.

5.2.1. Data processing and analysis

Data was analysed using custom made scripts in Matlab®. Raw sensor acceleration and angular rate signals (sampled at 110 Hz) were low pass filtered at 20 Hz with a zero phase 5th order Butterworth algorithm as explained in previous work (Morgado Ramírez et al., 2013c). Transmissibility of vibration was reported up to 8 Hz. Vibration transmitted through the thoracic spine was described through mean maximum transmissibility at frequency intervals ($0.5 \leq f \leq 2$, $2 < f \leq 4$, $4 < f \leq 6$ and $6 < f \leq 8$ Hz), maximum transmissibility at maximum acceleration PSD (maxT@maxPSD) and RMSa. Transmissibility of vertical vibration along the thoracic spine was estimated as the ratio of the power spectral density (PSD) of the output (T1) over the PSD of the input (T12) and from 0.5 to 8 Hz (Figure 5.2-1). Transmissibility was calculated for all physical

activities and for each group. Mean maximum transmissibility values with a 95% confidence interval were determined for each subject in order to study the frequencies at which the highest transmissibility of vibration was obtained. Maximum transmissibility at maximum acceleration PSD was determined for each subject. In order to measure transmitted vibration magnitude, RMSa was calculated for T1, T8 and T12. These calculations were made with vertical acceleration corrected for skin movement and sensor inclination for each trial.

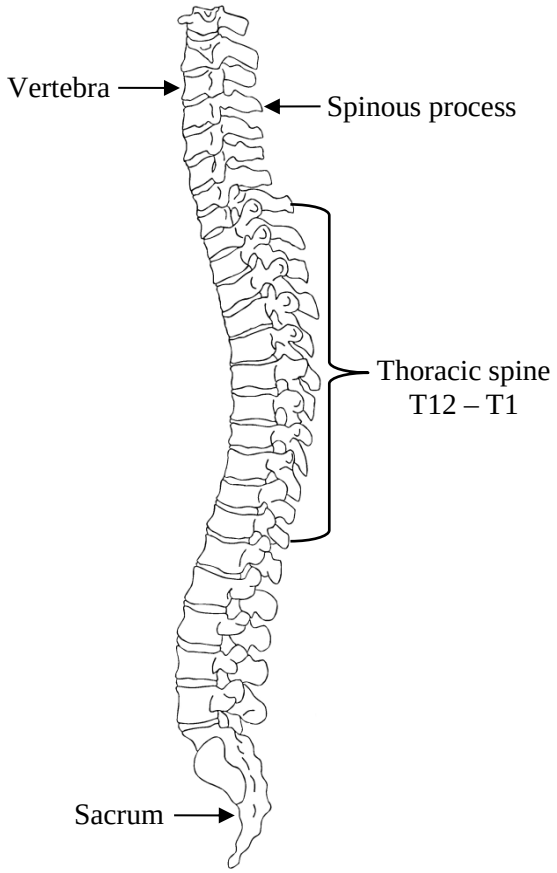

Figure 5.2-1 Transmissibility case based on normal thoracic spine curvature. First thoracic vertebra (T1), twelfth thoracic vertebra (T12)

5.2.2. Statistical analysis

Statistics were determined with IBM®SPSS® statistical software. All data was significantly non normal as determined by Kolmogorov-Smirnov tests and visual exploration through histograms against a normal curve, for each group and each physical activity. To test the hypotheses, differences in maximum transmissibility at frequency intervals, maxT@maxPSD and RMSa

between physical activities and between groups were determined through Kruskal-Wallis ANOVA with a 0.05 significance level. Differences in thoracic kyphosis between groups were also determined through Kruskal-Wallis. Post hoc tests were performed with a Bonferroni correction. To test the hypothesis that RMSa and maximum transmissibility at frequency bands are significantly different between lumbar and thoracic spines, significant differences were calculated through the Mann-Whitney U test for all physical activities and groups. A forward multiple regression analysis was performed to determine the variables that are significant contributors of maximum transmissibility. One regression model was obtained for each group employing independent variables such as thoracic kyphosis, RMSa, BMI, T-score, age, gender, and walking speed.

5.3. Results

5.3.1. Walking speed, skin correction factors and spine curvature

Mean walking speeds for each physical activity are presented in Table 4.3-1. Skin-sensor interface movement correction was done individually and for each sensor. Mean damping factors and natural frequencies utilized to correct for the skin-sensor interfaces movement are presented (Table 4.3-2). Not all subjects had the same quantity of sensors during trials due to technical problems or because the subject did not performed a particular trial. Details can be seen in Appendix B. The highest mean natural frequency was found for the skin-sensor interface of T8. Thoracic kyphosis was not significantly different across groups. Mean spine curvatures are presented in Figure 5.3-1.

Table 5.3-1 Correction fators for skin-sensor interfaces over the spine, all groups, mean (SD)

Group	Location	T1	T8	T12
YH	***fn* (Hz)**	13.123 (1.607)	19.981 (3.726)	16.452 (2.711)
	ζ	0.175 (0.083)	0.193 (0.12)	0.207 (0.127)
OH	***fn* (Hz)**	13.273 (2.526)	22.606 (5.296)	17.539 (3.898)
	ζ	0.297 (0.071)	0.391 (0.090)	0.405 (0.058)
OO	***fn* (Hz)**	12.290 (1.975)	19.419 (4.581)	15.853 (2.871)
	ζ	0.278 (0.072)	0.383 (0.105)	0.366 (0.072)

Natural frequency (f_n), damping ratio (ζ), young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), eighth thoracic vertebra (T8), twelfth thoracic vertebra (T12)

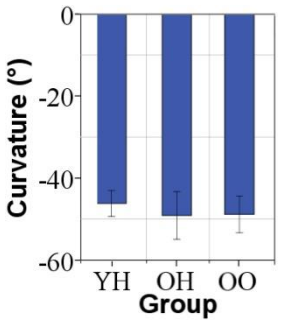

Figure 5.3-1 Thoracic kyphosis between groups.Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

5.3.2. RMS acceleration

Median RMSa with bars indicating where 95% of it was found for all groups and locations of the spine can be seen in Figure 5.3-2. Significant differences between physical activities are indicated by continuous lines. Significant differences between groups according to their BMD are indicated by the name of the groups followed by an asterisk. Ageing is generally associated with a decrease in the magnitude of vibration transmitted to the thoracic spine at T1. Osteoporosis is associated with minimal effects in vibration magnitude (Figure 5.3-2). At T8, ageing decreases the magnitude of vibration only during *w* while osteoporosis further decreases it. At the level of T12 the effects of osteoporosis is the opposite of that found at T8. Ageing decreases the magnitude of vibration at T12 during stair negotiation as explained for the lumbar spine. The OO spine received the same magnitude of vibration as the YH spine during *w* and *a* (at T8 and T12) and during *m* at T8 (Figure

5.3-2). The OH spine received the same magnitude of vibration as the YH spine at the level of T8 (during all physical activities except *m*) and at T12 during *w* and *m*. RMSa was also significantly affected by type of physical activity performed (Figure 5.3-2). For all locations over the spine and all groups, the magnitude of RMSa was significantly different between *a* and *d*, between *w* and *a*, between *m* and *d* and between *w* and *m* (except at T8 for the YH spine). For the YH group, *w* and *d* produced the same magnitude of RMSa only at level of T1. In contrast, this was observed at all spine levels for the OH group and at T1 for the OO group. Combined walking and turning and *a* produced the same magnitude of RMSa at T1 and T12 of the YH spine. Conversely, this was observed only at T8 and T12 for the OH spine and at T1 for the OO spine.

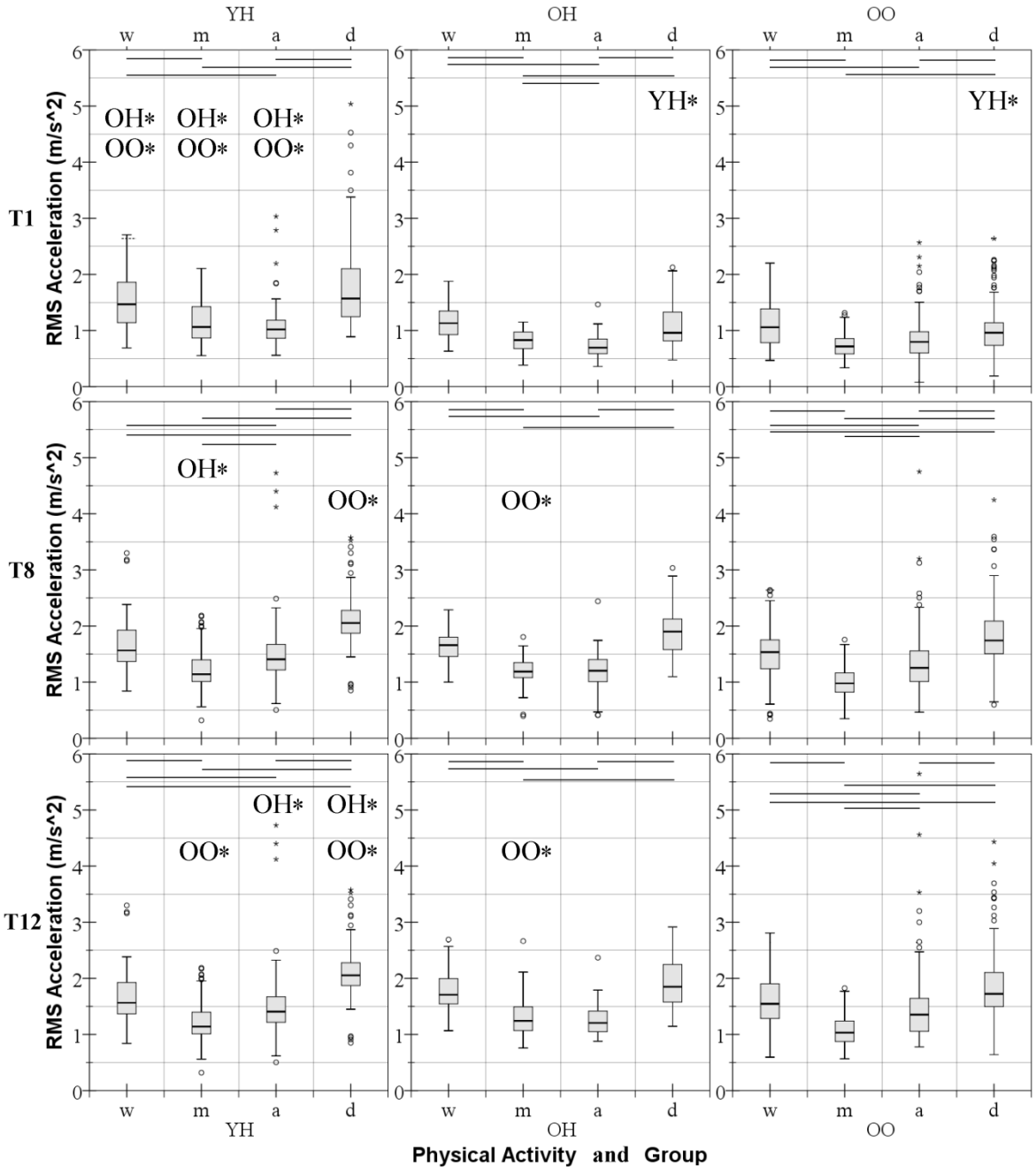

Figure 5.3-2 Root mean square (RMS) acceleration at T12, T8 and T1. Comparison between physical activities for all groups. ─ or * = significant difference. Young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d), first sacral vertebra (S1), twelfth thoracic vertebra (T12), eighth thoracic vertebra (T8)

5.3.3. Transmissibility overview

Mean transmissibility and acceleration PSD ((m/s2)2/Hz) are presented with 95% confidence intervals for comparison of the three groups during *w* (Figure 5.3-3). Only the YH group

transmissibility curve presented a clear different amplification zone (between 3.5 and 4.5 Hz) in comparison with the other two groups. All groups amplified vibration at very low frequency (0.5 to 1.25 Hz). It is also possible to see that at some frequencies the OH group had a greater attenuation than the OO group, yet both remained in the attenuation zone.

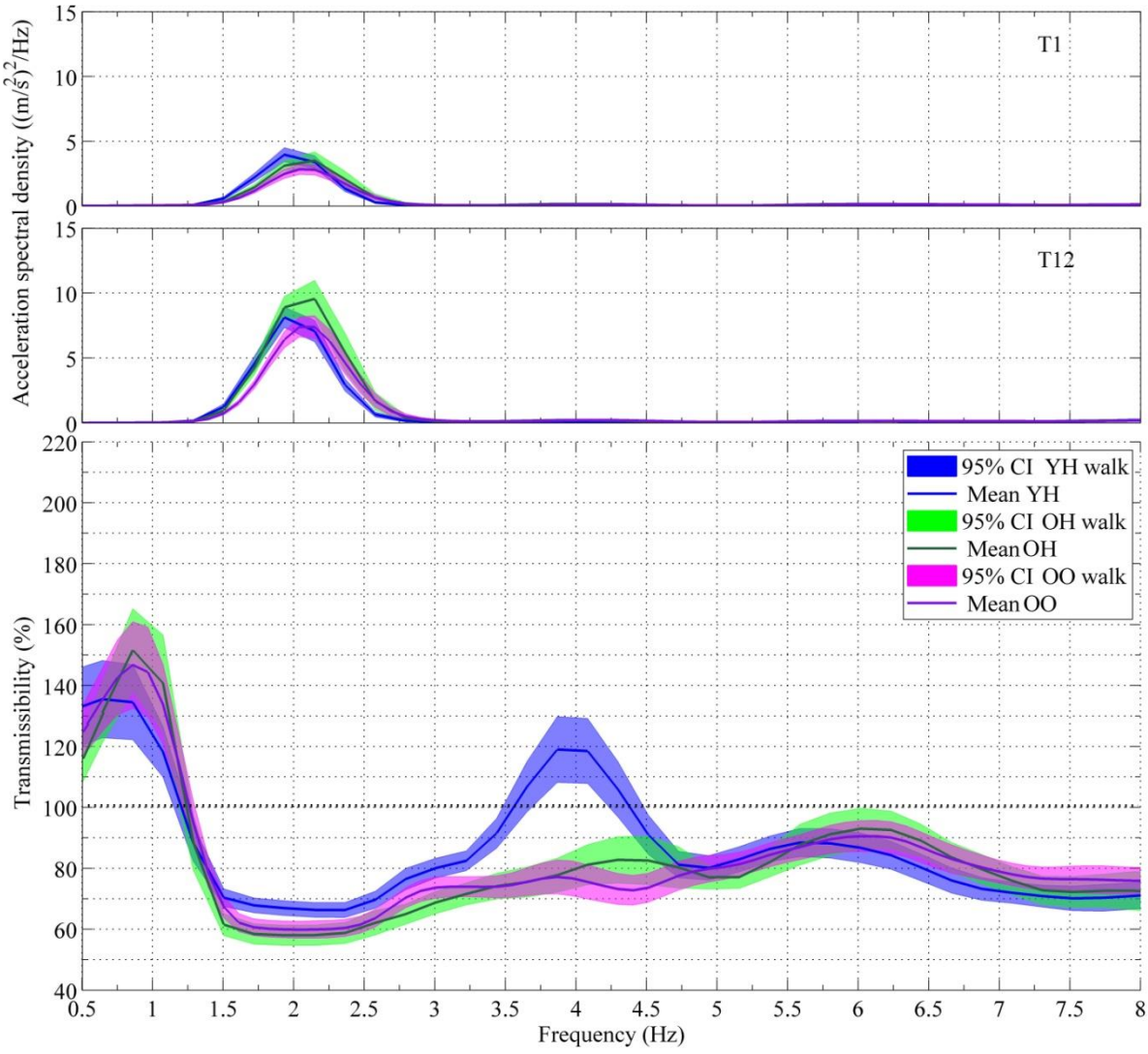

Figure 5.3-3 Input (T12) and output (T1) acceleration PSD and transmissibility, walking in a straight line during normal walking speed for all groups. First thoracic vertebra (T1), twelfth thoracic vertebra (T12)

Mean transmissibility and acceleration PSD $((m/s^2)^2/Hz)$ are presented with 95% confidence intervals for all physical activities for the YH thoracic spine (Figure 5.3-4). Vibration transmitted during *w* was amplified from 0.5 Hz to up to approximately 1.25 Hz to be later amplified again

between approximately 3.5 and 4.5 Hz. Combined walking and turning amplified only below approximately 1.25 Hz. Stair ascent and descent present less number of oscillations and mainly attenuation of vibration by transmitting less than 80%.

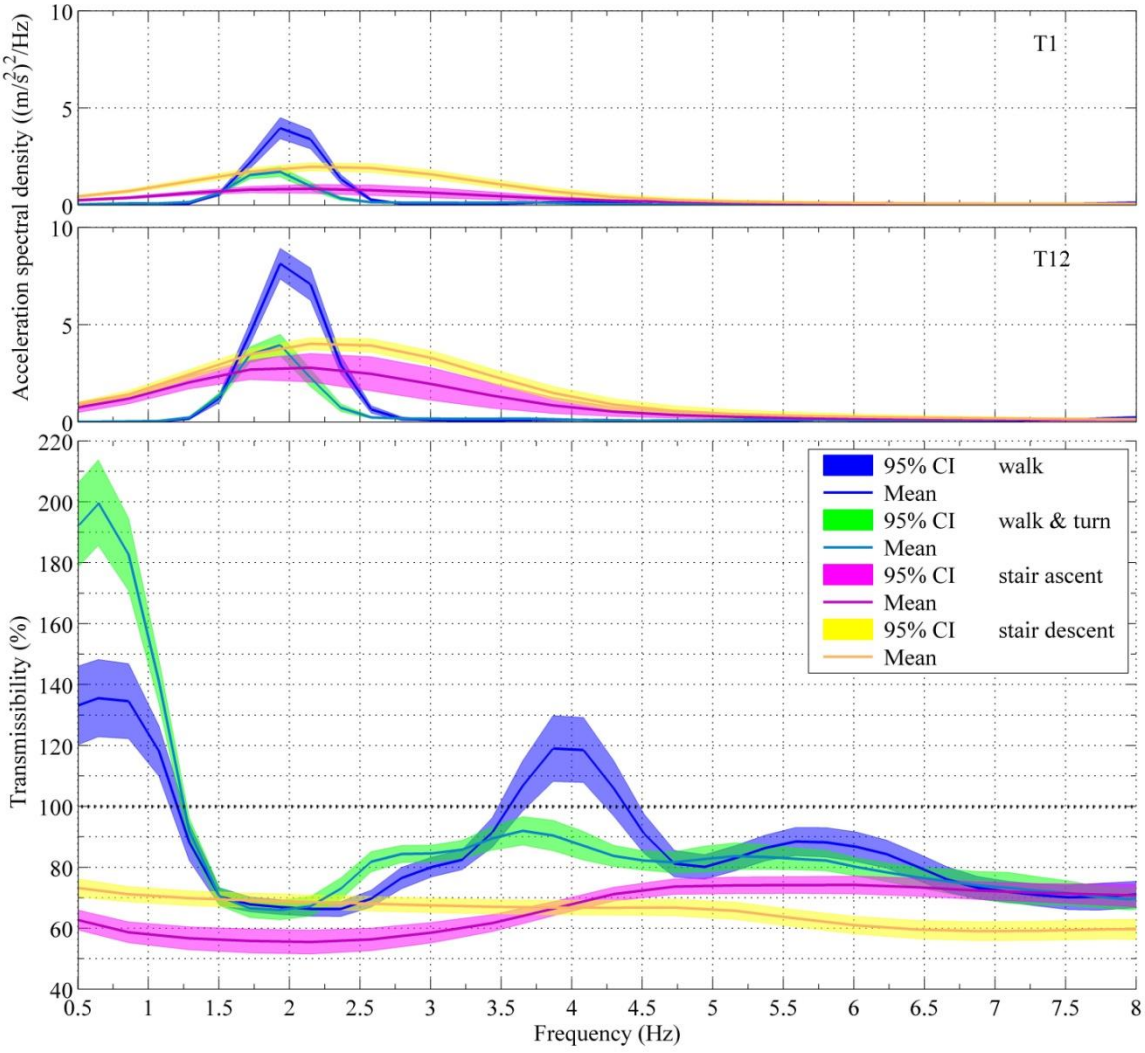

Figure 5.3-4 Input (T12) and output (T1) acceleration PSD and transmissibility during normal walking speed at all physical activities for the young and healthy group

5.3.4. Mean maximum transmissibility at frequency intervals

Mean maximum transmissibility with 95% confidence intervals were determined for 4 different frequency intervals: $0.5 \leq f \leq 2$, $2 < f \leq 4$, $4 < f \leq 6$ and $6 < f \leq 8$ Hz (Figure 5.3-5). A dotted reference line at the level of 100% transmissibility helps identify where attenuation and amplification of

vibration took place. Significant differences between groups according to their BMD are indicated by the name of the groups followed by an asterisk. Vibration transmissibility is attenuated except during walking at low frequency (<2Hz) regardless of ageing and osteoporosis (Figure 5.3-5). Ageing increases vibration attenuation during stair descent from 2 to 4 Hz. The three spines (YH, OH and OO) transmitted the same percentage of vibration during *a* and *m* ($0.5 \leq f \leq 2$ Hz) and during *w* ($0.5 \leq f \leq 2$ and $4 < f \leq 8$ Hz). Osteoporosis had a significant effect on mean maximum transmissibility at different frequency intervals and especially during *d*. The OH and OO spines transmitted significantly less vibration compared with the YH at all frequency intervals during *d* and from2 to 4 Hz during *m* and *w*. Similarly the OO spine transmitted significantly less vibration than the YH spine during *a* ($2 < f \leq 8$ Hz) and during *m* ($4 < f \leq 8$ Hz). Osteoporosis had minimal effect in vibration transmissibility. The OH spine transmitted significantly more vibration than the OO during *m* ($4 < f \leq 8$ Hz). During *a*, the healthy spines transmitted the same percentage of vibration at all frequency intervals. However, the OO spine transmitted significantly less vibration than the YH spine during *a* ($2 < f \leq 8$ Hz).

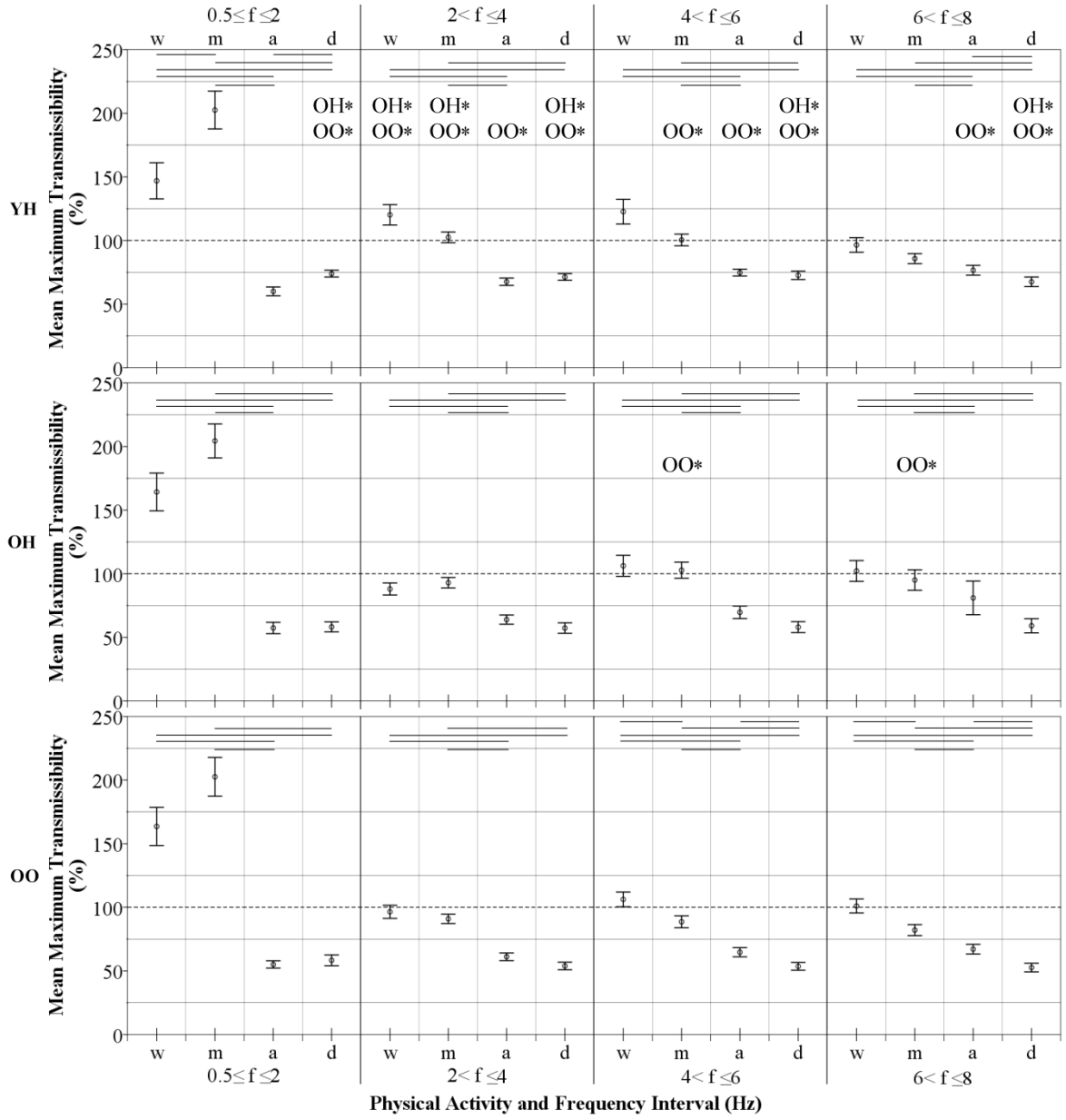

Figure 5.3-5 Mean maximum transmissibility from T12 to T1 at frequency intervals. ─ or * = significant difference. Dotted line= 100% transmissibility, attenuation below and amplification above it. Young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d), first sacral vertebra (S1), twelfth thoracic vertebra (T12)

Mean maximum transmissibility at the YH, OH and OO thoracic spines was significantly different between *w* and *a*, *w* and *d*, *m* and *d* and between *m* and *a* at all frequency intervals (Figure 5.3-5). For the YH spine, mean maximum transmissibility was significantly different between *a* and *d* for

the frequency intervals of 0.5≤ f ≤2 and 6< f ≤8 Hz only. Mean maximum transmissibility during *w* and *m* had were not significantly different from 2 to 8 Hz. Mean maximum transmissibility for the OO thoracic spine was significantly different between *w* and *m* as well as between *a* and *d* for the frequency intervals of 4< f ≤8 Hz. Overall the YH, OH and OO thoracic spines attenuated mean maximum transmissibility during *a* and *d* at all frequency intervals (Figure 5.3-5). The YH spine amplified vibration during *w* and *m* only between 0.5 and 6 Hz reaching a maximum of 148 ± 67% during *w* and 203 ± 72% during *m*. The OH spine reached the greatest amplification during *m* (204 ± 54%) followed by *w* (162 ± 60%) from 0.5 to 2 Hz. The OO thoracic spine reached the greatest amplification during *m* (202 ± 81%) from 0.5 to 2 Hz whereas the least maximum transmissibility was observed during *d* from 6 to 8 Hz (52 ± 18%).

5.3.5. Mean maximum transmissibility at maximum acceleration PSD

All physical activities had the same distribution for maxT@maxPSD for the OH and OO spines (Figure 5.3-6). Significantly different magnitudes of maxT@maxPSD were found between m and a as well as between a and d for the YH group. The lowest transmissibility (54 ± 10 %) at maximum acceleration PSD for the YH lumbar spine was observed during a. Conversely, the maximum transmissibility was observed during w (70 ± 15%). The lowest transmissibility observed for OH and OO thoracic spines were of 53 ± 14% (a) and 53 ± 16% (d) respectively. The highest transmissibility observed for OH and OO thoracic spines were of 59 ± 14% and 59 ± 15% both during w.

Significant differences between groups according to their BMD are indicated by the name of the groups followed by an asterisk (Figure 5.3-6). Significant differences between groups based on maxT@maxPSD were observed during d only. The YH spine transmitted significantly more vibration than the OH and OO spines during d. Attenuation of vibration is indicated by maxT@maxPSD for the thoracic spine.

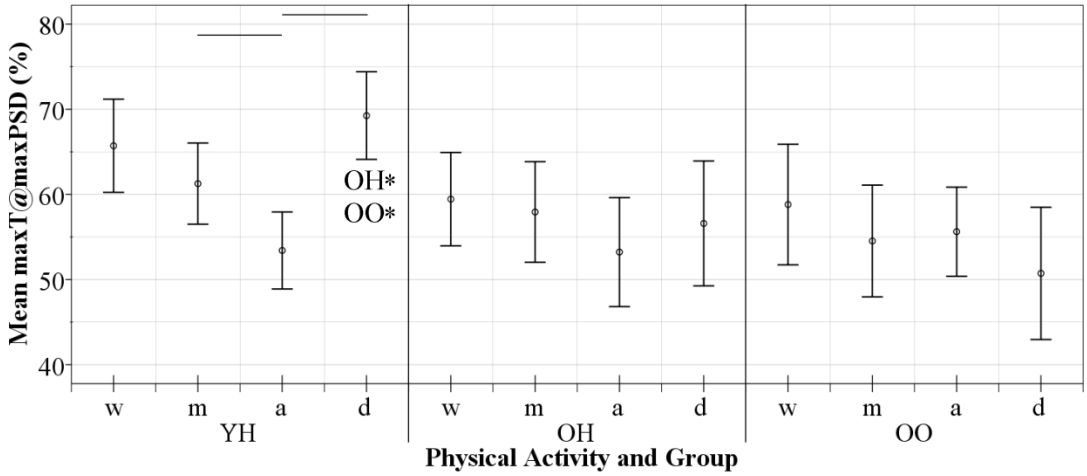

Figure 5.3-6 Maximum transmissibility at maximum acceleration PSD (maxT@maxPSD) during normal walking speed for the thoracic spine. ─ or * = significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

5.3.6. Differences between lumbar and thoracic spines

Significant differences in mean maximum transmissibility were found between the lumbar and thoracic spines (Figure 5.3-7). Relative percentage differences in mean transmissibility from lumbar to thoracic spine were calculated. For the frequency interval of $2< f \leq 8$ Hz the OH and OO thoracic spines transmitted significantly less vibration (between 16% and 60% less) compared with the lumbar spines during w and m. During a and d, thoracic mean maximum transmissibility was significantly lower (20% and 60% less) for all groups and for the entire frequency interval studied. The YH thoracic and lumbar spines transmitted the same percentage of vibration during w from 0.5 to 6 Hz and during m from 4 to 6 Hz, in contrast with the older spines (OH and OO) where significant differences between spine sections were observed. During m (from 0.5 to 2 Hz) the thoracic spine transmitted significantly more vibration than the lumbar spine for all groups (between 44% and 56%). Similarly, the OH and OO thoracic spine transmitted significantly more vibration (20 to 28% greater) than the lumbar spine during w from 0.5 to 2 Hz. The OO thoracic spine transmitted the same amount of vibration as the lumbar spine during w from 0.5 to 2 Hz, the

same was seen for the YH thoracic spine but at the additional frequency interval of $4< f \leq 6$ Hz (Figure 5.3-7).

Significant differences in the mean frequencies (f), at which mean maximum transmissibility was found, between the lumbar and thoracic spines were found for specific frequency intervals and physical activities (Figure 5.3-8). Mean maximum transmissibility at the thoracic spine was found at significantly lower frequencies in comparison with the lumbar spine at the frequency intervals of $0.5\leq f \leq 2$ and $6< f \leq 8$ Hz for all groups and physical activities. Mean maximum transmissibility at the thoracic spine of all groups was found at significantly greater frequencies (4 to 52% greater), in comparison with the lumbar spine, during all physical activities from 2 and 4 Hz (except during d). Mean maximum thoracic transmissibility of the OH thoracic spine was found at significantly lower frequencies (4 to 8% lower) than the lumbar one during a and d from 4 to 6 Hz.

Maximum transmissibility at maximum acceleration PSD was significantly lower at the thoracic spine (between 35% and 59% less) for all groups and during all physical activities (Appendix C3). Thus physical activities do not have any effect on the differences between lumbar and thoracic spines because this difference is the same across the physical activities studied. Same applies for the frequencies at which maximum transmissibility at maximum acceleration PSD was found (Appendix C5).

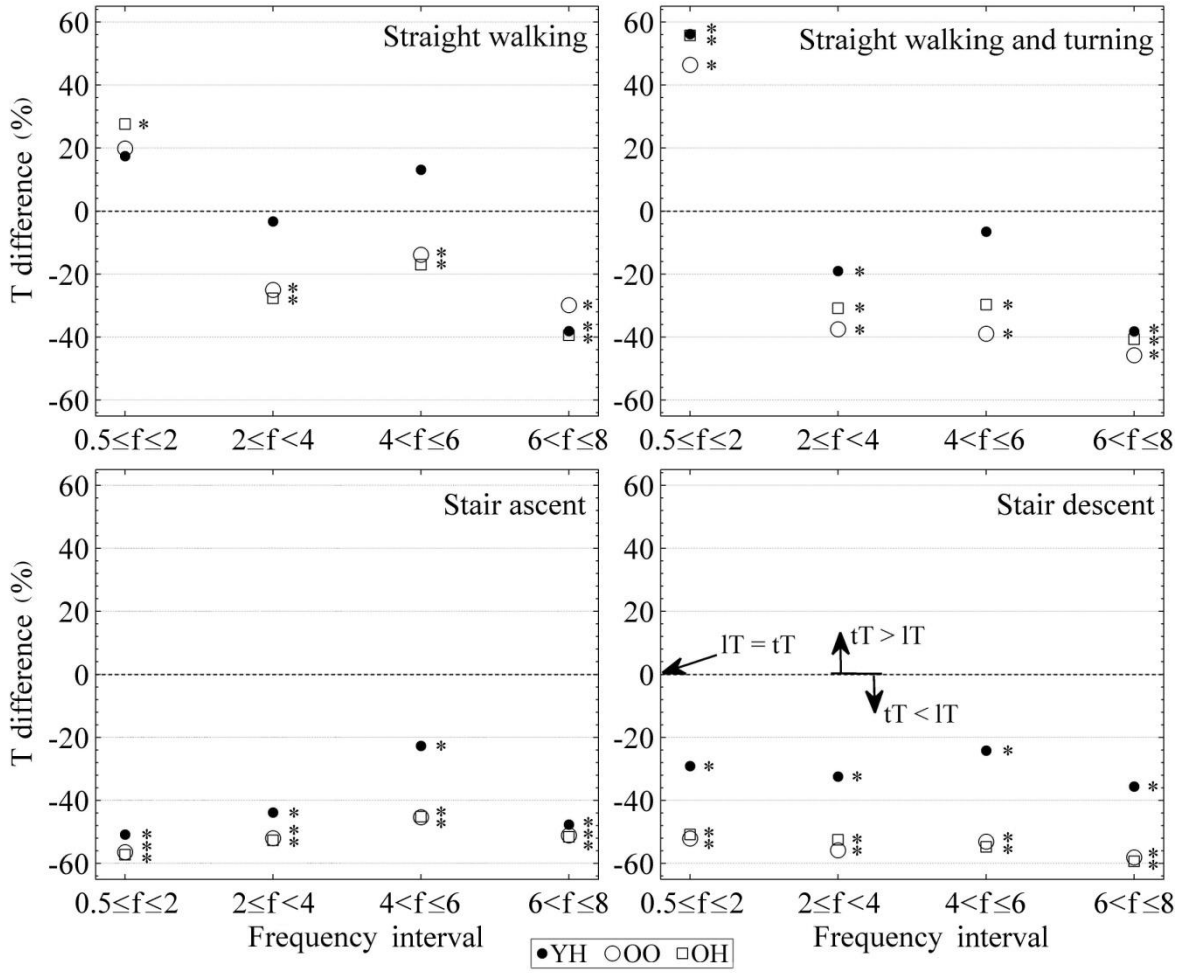

Figure 5.3-7 Percentage difference in mean maximum transmissibility (T) at frequency intervals between the thoracic spine (t) and the lumbar spine (l). *=significant difference in T between l and t. Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

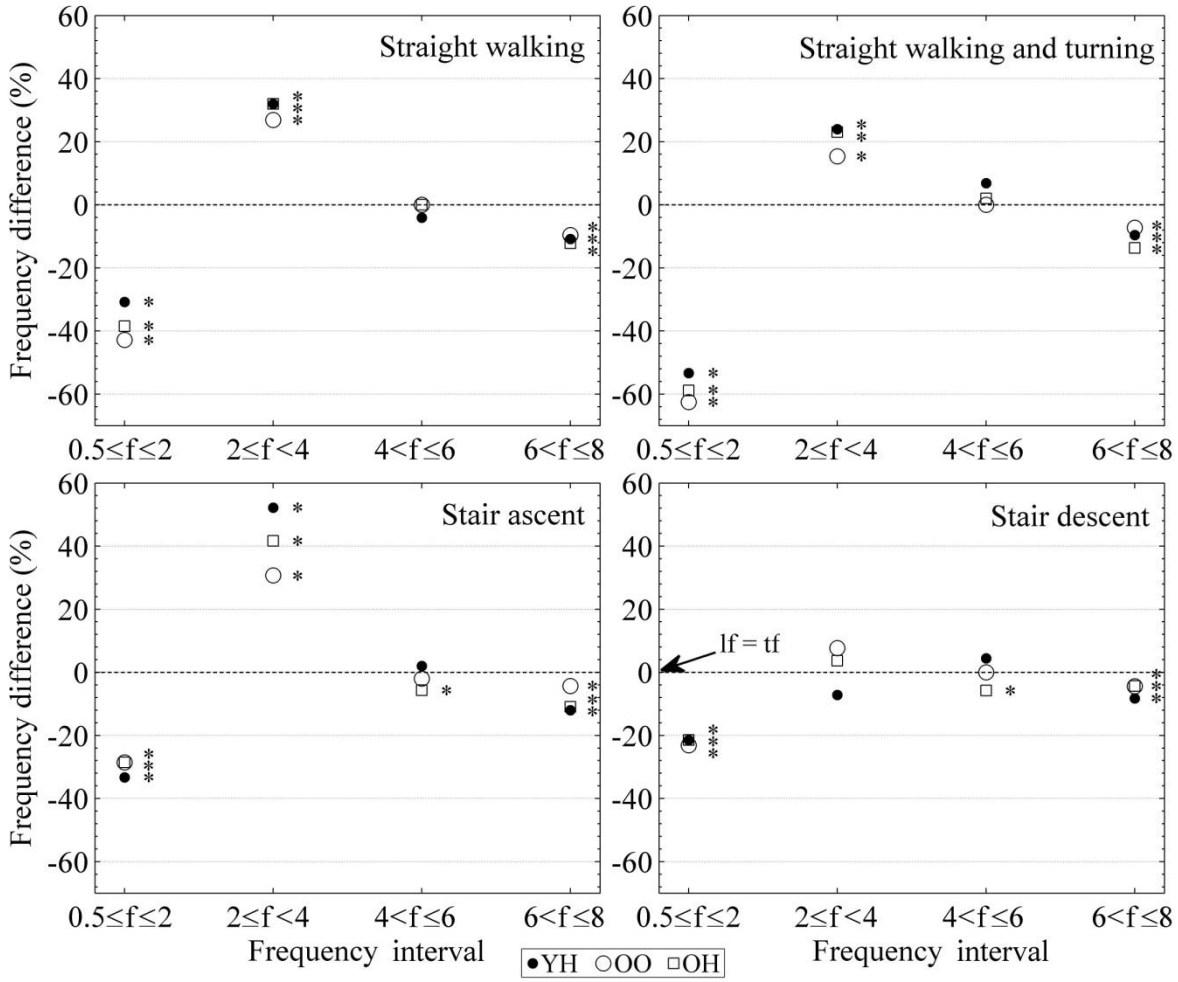

Figure 5.3-8 Percentage difference in the frequency (f) at which mean maximum vibration transmissibility at frequency intervals was found between the thoracic spine (t) and the lumbar spine (l). *=significant difference in f between l and t. Young and healthy (YH), older healthy (OH), older osteoporotic (OO)

5.3.7. Transmissibility predictors

Thoracic kyphosis is an important predictor of transmissibility for all groups and significant for the YH had OO groups (Table 5.3-2). RMSa at T1 was the most important predictor for the transmissibility of the thoracic spine for all groups. 64% of the transmissibility variability was explained for the OO group and 40% and 47% for the YH and OH groups respectively. Gender was considered as a predictor for the young and healthy group only due the equivalent distribution of gender among this group only. The specific transmissibility variability explained by thoracic kyphosis for each group was between 1.5% and 4.7%.

Table 5.3-2 Models to predict mean maximum transmissibility at maximum acceleration PSD for the thoracic spine during normal walking speed

Group	r^2	Residual	Variability explained	Model Term	Coefficient	Significance	Importance
YH	0.402	1.186	40.2%	Intercept	1.044	0.000	
				RMSa at T1	0.096	0.000	0.134
			4.7%	Thoracic Kyphosis	0.003	0.001	0.127
				Female	0.053	0.002	0.126
				BMI	-0.011	0.004	0.125
				Age	-0.006	0.020	0.123
				RMSa at T12	-0.066	0.026	0.122
				WS	0.039	0.036	0.122
				RMSa at T8	0.048	0.143	0.120
OH	0.470	0.975	47%	Intercept	0.289	0.012	
				RMSa at T1	0.457	0.000	0.315
				RMSa at T12	-0.218	0.000	0.253
				Age	0.004	0.008	0.221
			1.5%	Thoracic Kyphosis	0.001	0.109	0.212
OO	0.640	1.360	64%	Intercept	0.853	0.000	
				RMSa at T1	0.434	0.000	0.201
			2.9%	Thoracic Kyphosis	0.002	0.000	0.118
				RMSa at T12	-0.121	0.001	0.117
				WS	-0.048	0.005	0.115
				T-score	0.025	0.009	0.114
				RMSa at T8	-0.076	0.036	0.112
				BMI	-0.006	0.053	0.112
				RMSa at S1	-0.042	0.090	0.111

Root mean square acceleration (RMSa), young and healthy (YH), older healthy (OH), older osteoporotic (OO), body mass index (BMI), first sacral vertebra (S1), twelfth thoracic vertebra (T12), eighth thoracic vertebra (T8), first thoracic vertebra (T1), walking speed (WS)

5.4. Discussion

Results supported the hypothesis that ageing has a significant effect on the magnitude and percentage of vibration transmitted to T8 and T12. Similarly, osteoporosis has a significant effect

but different from that of ageing. Different physical activities produce significantly different effects on vibration transmitted to T8 and T12 where vertebral fractures are common. The differences in transmissibility seen between lumbar and thoracic spines are significant. Thoracic kyphosis has a small but significant and important contribution to the vibration transmitted.

The thoracic spine attenuates vibration regardless of ageing and osteoporosis except during walking at low frequency (<2Hz) (Figure 5.3-5). A previous study applied thrusts to the spinous processes of the lumbar and thoracic spines to quantify the transmissibility with transient vibration (Keller et al., 2000). Keller et al. (2000) also observed that the lumbar region exhibits higher stiffness when compared with the thoracic spine. Other studies have measured transmissibility of healthy spines during walking from S1 to the second thoracic vertebra (T2) (Smeathers, 1989b) and from sacrum to T2 (Smeathers, 1989a) from 1 to 40 Hz. Comparison with these previous studies is difficult since transmissibility was measured in a single subject and in a section of the spine that included the lumbar and thoracic spines. It is not possible to separate the response of the thoracic section. Vibration transmissibility of OH and OO thoracic spines during physical activity has not been measured before.

Ageing increased vibration attenuation during stair descent from 2 to 4 Hz while osteoporosis had no effect generally (Figure 5.3-5). Ageing is associated with potential conflicting consequences. Ageing increases vibration attenuation which may either offer protection or remove the mechanical stimulus necessary to stimulate bone in the thoracic region. The thoracic spine presented frequency intervals during which older spines transmitted the same percentage of vibration during walking (straight and with turning) and during stair ascent regardless of osteoporosis (Figure 5.3-5). Where this effect was not observed, transmissibility of OH and OO spines was either at the same percentage as the YH or below it (at all frequencies and physical activities). This behaviour presents a challenge for clinicians willing to stimulate a thoracic spine with current vertebral fractures or with high risk of fracture as with osteoporosis. Given that ageing is associated with less transmissibility in the thoracic spine, further research is necessary to understand if increasing the

stimulus delivered to the spine (for example, through exercise) can overcome the effects of ageing and if that stimulus will be beneficial at all.

It is important to note that the thoracic spine presented vibration amplification only during walking (straight and with turning) (Figure 5.3-5). This suggests that there are other factors that determine vibration transmissibility capacity of the thoracic spine than just material properties of bone. On one hand, the amplification seen may be triggered by the stiffer tissue due to ageing and by active muscular contraction at very low frequencies (Feltham et al., 2006, Willigenburg et al., 2010, Huang and Griffin, 2006, Orkoula et al., 2012, Yerramshetty and Akkus, 2008). On the other hand, the general attenuation of the thoracic spine may be due to the effect of the thoracic cage, the thoracic kyphosis and to the smaller thoracic vertebrae in comparison with lumbar ones (Watkins et al., 2005, 2007, Tortora and Grabowski, 2003). It has been suggested that voluntary periodic body movement reduces the stiffness of the body when exposed to low frequency vibration (Huang and Griffin, 2006). A reduction of stiffness means that spinal damping is increased. Transmission of vibration through the spine is dependent on tissue properties as well as structure. Therefore spinal tissue damping properties, curvature and various types of tissue deterioration may affect the vibration transmissibility capability of the thoracic section.

When employing maxT@maxPSD, the thoracic spine preserved its attenuation tendency (Figure 5.3-6). The greatest maximum transmissibility was produced only during straight walking while the smallest maximum transmissibility was seen during stair ascent and descent. The YH thoracic spine was the only one that was significantly affected by the type of activity performed. The transmissibility capacity of the older healthy and osteoporotic thoracic spines was not sensitive to different physical activities when employing this measurement.

Ageing is generally associated with a decrease in the magnitude of vibration. Osteoporosis is generally associated with both increment and decrement of attenuation at different levels of the thoracic spine (Figure 5.3-2). The greatest vibration magnitudes are observed during walking and descending stairs. Walking may be recommended while descending stairs should consider factors

such as the risk of falling. But it still remains to determine if that vibration magnitude produced during walking is beneficial for bone growth or not. The measurement of RMSa has suggested that the OH and OO spines receive a similar magnitude of vibration as the YH spine at T8 and T12 during specific physical activities (Figure 5.3-2). The significance of these results is that the musculoskeletal system of the thoracic spine may offer no protection against vertebral fractures for people with osteoporosis. It is concerning that vertebral fractures may occur during physical activity without serious symptoms (Lems, 2007). Individuals with BMD not classified as osteoporosis also present vertebral fractures (Kanis et al., 2008). The relationship between RMSa dose and vertebral fracture cannot be determined from this study. Future research should aim to identify those physical activities which expose specific locations of the thoracic spine to vibration magnitudes that may be classified as harmful (increase risk of fracture) or beneficial (either promote bone growth or decrease bone loss). This represents a major challenge for clinicians since it has been suggested that only physical activities such as jumping and running can produce an increment in BMD by providing peak accelerations greater than 52.9 m/s^2 (Johnell and Kanis, 2006). Future research needs to weight the risk of fracture against bone strength gain.

In general, differences in transmissibility between the thoracic and lumbar spines spine are statistically significant. Physical activity caused significant effects at the thoracic spine more often than at the lumbar spine. It is also important to note that for certain frequencies the attenuation and amplification capabilities were equal between lumbar and thoracic spines (Figure 5.3-7). The YH lumbar and thoracic spines transmitted the same percentage of vibration from 0.5 to 6 Hz during straight walking. In contrast, an OO spine tested under the same conditions suggests that this equality of transmission between lumbar and thoracic spines is reduced to frequencies between 2 and 6 Hz. This further suggests that there are other factors that determine vibration transmissibility of the different sections of the spine. These factors are delimited by diverse anatomical, functional and viscoelastic material properties. It is suggested that prescribed physical activity as part of a healthy lifestyle or as a treatment for osteoporosis should consider the differences in the mechanical response between lumbar and thoracic spine.

5.4.1. Transmissibility predictors

Root mean square acceleration, thoracic kyphosis, BMI, T-score, gender and walking speed proved to be important and significant variables that could explain from 40.2% up to 64% of vibration transmitted through the thoracic spine during physical activities. Thoracic kyphosis has a positive, important and significant contribution to transmissibility, even when no significant differences in thoracic curvature between groups are present. T-score has a positive effect on transmissibility for the older osteoporotic group (Table 5.3-2). Large transmissibility variability seen between subjects may account for the unexplained transmissibility variability. High intra subject transmissibility variability has been reported previously (Kitazaki and Griffin, 1995). Other variables that could explain the remaining transmissibility variability may be related to geometric and material properties that cannot be measured in vivo.

The spine is made from several viscoelastic materials, thus the implication of osteoporosis on its dynamic response (in terms of damping and stiffness) can only be truly understood when exposed to realistic in vivo loading. Before this research study, it was not possible to tell with certainty the effect of osteoporosis, age and different physical activities on vibration transmission of the thoracic spine. There is considerable evidence that RMSa measurement is capable of detecting peculiarities during specific physical activities and at levels of the thoracic spine where fractures are common. This offers an exciting tool for research and development as well as for clinical use. For instance, the objective will be to identify exercise intensity and frequency currently causing sudden vertebral fractures in populations with and without spinal osteoporosis. Before an exercise routine can be prescribed, it is necessary to understand the magnitude and percentage of vibration that would cause a vertebral fracture. The technique presented in this study provides an unprecedented tool for clinical researchers. Vibration transmission could help study if current physical activities recommended as a treatment for osteoporosis safely and effectively stimulate levels of the thoracic spines known to fracture. Future research will potentially use this technique along with bone medical imaging in order to determine an accurate model for the prediction of vibration transmission and its implication for bone health.

5.4.2. Limitations

The calculation of vibration transmitted through the spine is an approximation due to the skin-sensor interface movement correction being a mathematical model (Kim et al., 1993, Ziegert and Lewis, 1978). The skin correction method requires subjects to be in a healthy weight. One limitation of RMSa is that it provides the magnitude of vibration at all frequencies up to 20 Hz. However, animal studies have demonstrated that peak bone loading occurs at relatively low frequency (1-3Hz) (Thompson et al., 2012). Similarly, previous studies have reported that low frequency mechanical loading is effective simulating bone but no agreement has been achieved regarding the magnitude of that stimulation (Turner et al., 1994, Burr et al., 2002). Therefore it is believed that RMSa measurement provides a tool to characterize the intensity of physical activity at frequencies which have demonstrated their potential to stimulate bone growth. Another limitation is that the physical activities tested here are an incomplete range of everyday physical activity. The number of sensors available for this study was also a limitation, future research may benefit from attaching smaller inertial sensors over each vertebrae for a more detailed profile of vibration transmission and magnitude.

A greater sample of the population is recommended to account for variability between subjects, to include variables such as risk of fracture, ethnicity, alcohol and tobacco consumption, family history of osteoporosis and use of pharmacological treatments. The T-scores were determined through QUS which provide a sufficient but not extensive overview of the general health of the skeleton. The results of this study provide independent observations on the differences in vibration transmission due to osteoporosis, age and physical activity. It is not possible, from these results, to determine relationships or whether a difference is driven by a third unmeasured variable.

5.5. Conclusion

The thoracic spine attenuated most vibration produced during gait. Vertebrae known to often fracture in older individuals (with and without osteoporosis) experience the same vibration transmission as a young and healthy individual. It is suggested that ageing has greater effects on the mechanical response of the thoracic spine when compared with those effects caused by

osteoporosis during physical activity. Differences in vibration transmission between lumbar and thoracic sections were seen due to osteoporosis. Thoracic kyphosis is an important and significant determinant of vibration transmission. Further research should employ this technique in randomized controlled trials to identify the intensity and types of physical activities that significantly increase the risk of vertebral fractures in people with osteopenia and osteoporosis while taking into account the different mechanical response of the lumbar and thoracic sections.

5.6. Key Findings

1. The thoracic spine generally attenuates vibration transmitted during physical activities at a self selected normal walking speed
2. Ageing is associated with potential conflicting consequences: it increases vibration attenuation which may either offer protection or remove the mechanical stimulus necessary to stimulate bone in the thoracic region.
3. Ageing is generally associated with a decrease in the magnitude of vibration and osteoporosis with both increment and decrement of attenuation at different levels of the thoracic spine.
4. Individuals with significantly different BMD may be receiving the same magnitude of vibration to levels of the thoracic spine prompt to fracture with osteoporosis during daily physical activities.
5. It is suggested that prescribed physical activity as part of a healthy lifestyle or as a treatment for osteoporosis should consider that the differences in the mechanical response between lumbar and thoracic spine.
6. Osteoporosis and ageing significantly affect the differences in transmissibility of vibration between the lumbar and thoracic spines.
7. A small percentage of vibration transmission of the thoracic spine is determined by its curvature during gait.

CHAPTER 6 Effect of Fast Walking on Vibration Transmission through the Spine

6.1. Introduction

Physical activity is a non pharmacological complementary treatment that produces a small improvement in spinal BMD (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Cheung and Giangregorio, 2012). Recommended physical activities for people with spinal osteoporosis are sparse (walking, volleyball, Tai Chi, aerobics, strength training and a combination of physical activities). Brisk walking has also been recommended to improve bone health (Winter-Stone, 2005, NOF, 2010, Van Norman, 2010, NOF, 2012), while other studies have found no effect on spinal BMD (Martyn-St James and Carroll, 2008, Schmitt et al., 2009). Brisk walking has been defined as a physical activity of moderate intensity normally at speeds between 1.6 m/s and 1.8 m/s (Murphy and Hardman, 1998). It has also been defined as a comfortable pace that is faster than normal walking speed while never causing shortness of breath (Ebrahim et al., 1997).

It is not known how often and how much exercise is optimal for people with osteoporosis to respond safely and positively to exercise (Hamilton et al., 2010a, Cheung and Giangregorio, 2012, Kelley et al., 2013). There is a lack of research based evidence of the threshold at which physical activity would either improve bone health or further deteriorate already osteoporotic bone towards fracture (Kohrt et al., 2004, Rittweger, 2006, Hamilton et al., 2010a, Gremeaux et al., 2012, Cheung and Giangregorio, 2012).

Bone responds to mechanical stimulation in the form of vibration (Skerry, 2008, Chen et al., 2010) and the way this vibration is transmitted through the bone depends on its material and structural properties (Keller et al., 2000, Kawchuk et al., 2009, Bediz et al., 2010, Bhattacharya et al., 2010) as well as on the magnitude and frequency of that vibration (Mansfield, 2005a). Heel strikes during walking produce vibration that is transmitted through the body (Cappozzo, 1982, Collins and Whittle, 1989). Vibration transmitted through the body can be calculated as the ratio of the

vibration measured between two points, is a function of frequency and has no units (Mansfield, 2005a). When transmissibility it is greater than 100% vibration is amplified (Mansfield, 2005a). Transmissibility through the spine has been measured previously during physical activity but during self selected normal walking speed only (Morgado Ramírez et al., 2013a, Morgado Ramírez et al., 2013b). However the effect of fast walking on the magnitude and percentage of vibration transmitted through the spine is not known.

It was hypothesized that vibration transmission by the lumbar and thoracic spines is significantly affected by walking speed.

6.2. Methods

Participants and measurements were the same as in previous two chapters (Table 4.2-1), except that in this study fast walking speed was tested instead of normal walking speed. Participants were divided into three groups: young and healthy (YH), older healthy (OH) and older with osteopenia and osteoporosis (OO) (Table 4.2-1).

Four inertial sensors were used, one over the first sacral vertebra (S1) and three more over the spinous processes of the twelfth (T12), eighth (T8) and first (T1) thoracic vertebrae (Figure 4.2-2). First, participants were asked to ascend (a) and descend stairs (d) consisting of 15 steps of normal height and 1.19 m wide with a continuous hand rail on both sides. Secondly to perform walking along a straight line (w) and lastly through a path consisting of combined turning and walking (m). All physical activities were performed three times at fast (FWS) walking speed. Subjects were encouraged to achieve their fast walking speed by asking them to walk as fast as they safely and comfortably could without running (therefore preventing shortness of breath). A rest was given between trials to prevent fatigue.

6.2.1. Data processing and analysis

All data was processed using Matlab®. Acceleration was corrected for the inclination of the sensor to the vertical in the time spectrum and for skin movement in the frequency spectrum. Power spectral density (PSD) of global acceleration corrected for skin movement and low pass filtered at

20 Hz was calculated. Transmissibility of vertical vibration along the lumbar and thoracic spine was estimated as the ratio of the power spectral density (PSD) of the output (T12 for lumbar and T1 for thoracic spine) over the PSD of the input (S1 for lumbar and T12 for thoracic spine) and over the frequency interval of 0.5 to 8 Hz.

In order to measure the intensity of vibration transmitted to each location over the spine RMSa was calculated (Mansfield, 2005a). These calculations were made with vertical acceleration corrected for skin movement and sensor inclination at each location of the spine (T1, T8, T12 and S1).

Transmissibility was calculated for all physical activities, for each walking speed and for each group (according to their bone health). Mean curves with 95% confidence intervals for transmissibility was employed as a graphical method to present transmissibility general tendency and the effect of walking speed. Maximum transmissibility was determined to observe the frequencies at which the highest amplification of vibration was obtained for each subject. Mean maximum transmissibility values with a 95% confidence interval were determined for 4 different frequency intervals: $0.5 \leq f \leq 2$, $2 < f \leq 4$, $4 < f \leq 6$ and $6 < f \leq 8$ Hz. These mean maximum transmissibility values were compared between walking speeds for each frequency interval.

Maximum transmissibility at maximum spectral density (maxT@maxPSD) of the outputs (T1, T8, and T12) was calculated for each subject, for lumbar and thoracic spines, for all physical activities and walking speeds. These maximum transmissibilities at maximum spectral density of the output were used for multiple regression analysis.

6.2.2. Statistical analysis

Statistics were determined with IBM® SPSS® Statistics software. All data was considered non parametric due to the Kolmogorov-Smirnov test results and due to the relatively small sample size for each group. To test the hypotheses that mean maximum transmissibility at frequency intervals and RMSa are significantly different between physical activities and between groups when walking fast, Kruskal-Wallis ANOVA was employed with a 0.05 significance level. Post hoc tests were performed with a Bonferroni correction. To test the hypothesis that mean maximum

transmissibility at frequency intervals and RMSa are significantly different between walking at a normal speed and fast walking, significant differences were calculated through the Mann-Whitney U test for all physical activities and groups. Previously reported vibration magnitude and transmissibility at lumbar and thoracic spines during self selected normal walking speed (NWS) (Figure 4.2-5 and Figure 5.3-5) was compared with that measured during FWS in this study. Mann Whitney U tests were also employed to test the hypothesis that transmissibility measured during FWS was different between lumbar and thoracic spines. A significance level of 0.05 was chosen to delineate a statistically significant result. In order to determine the variables that are significant contributors of maximum transmissibility of vertical vibration at maximum spectral density, a forward multiple regression analysis was performed.

6.3. Results

6.3.1. Walking speed

All subjects successfully walked at a significantly higher walking speed compared with normal self selected walking speed. Differences in walking speed between groups were also found (Table 6.3-1). Maximum walking speed of the YH group was significantly different to that of the OO group for all physical activities. It also was significantly different between YH and OH groups during *m*. Self selected walking speed was the same as the OH and OO groups during *w* and during *a* and *d*. Self selected walking speed was significantly different between YH and OO as well as between OO and OH during *m*.

Table 6.3-1 Walking speeds for all physical activities and groups

Group	(m/s)	Physical activity							
		w		m		a		d	
YH	NWS	1.659 (0.199)		1.337 (0.198)	OO*	0.584 (0.077)		0.688 (0.107)	
	FWS	2.502 (0.278)	OO*	1.943 (0.230)	OH* OO*	0.949 (0.189)	OO*	1.147 (0.224)	OH* OO*
OO	NWS	1.625 (0.264)		1.192 (0.196)	OH*	0.591 (0.144)		0.668 (0.179)	
	FWS	2.181 (0.316)		1.582 (0.266)		0.840 (0.185)		0.937 (0.233)	
OH	NWS	1.752 (0.216)		1.325 (0.226)		0.590 (0.060)		0.697 (0.148)	
	FWS	2.330 (0.312)		1.787 (0.305)		0.860 (0.140)		0.965 (0.196)	

Mean (SD), * = significant difference between groups, young and healthy (YH), older healthy (OH), older osteoporotic (OO), normal walking speed (NWS), fast walking speed (FWS),straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

6.3.2. Lumbar spine

An example of mean transmissibility curves and acceleration PSD $((m/s^2)^2/Hz)$ is presented with 95% confidence intervals for *m* at NWS and FWS for the YH group (Figure 6.3-1). Maximum acceleration PSD at FWS is more than two times greater than that transmitted during NWS (Figure 6.3-1). During FWS the lumbar spine conserved its amplification tendency. Maximum transmissibility is approximately 10% greater than that found at NWS and found at a greater frequency. Transmissibility tendency at other frequencies seem to have a different tendency compared to the one found during NWS. For example, at approximately 5.5 Hz maximum transmissibility at NWS was approximately 65% and at FWS was approximately 90%.

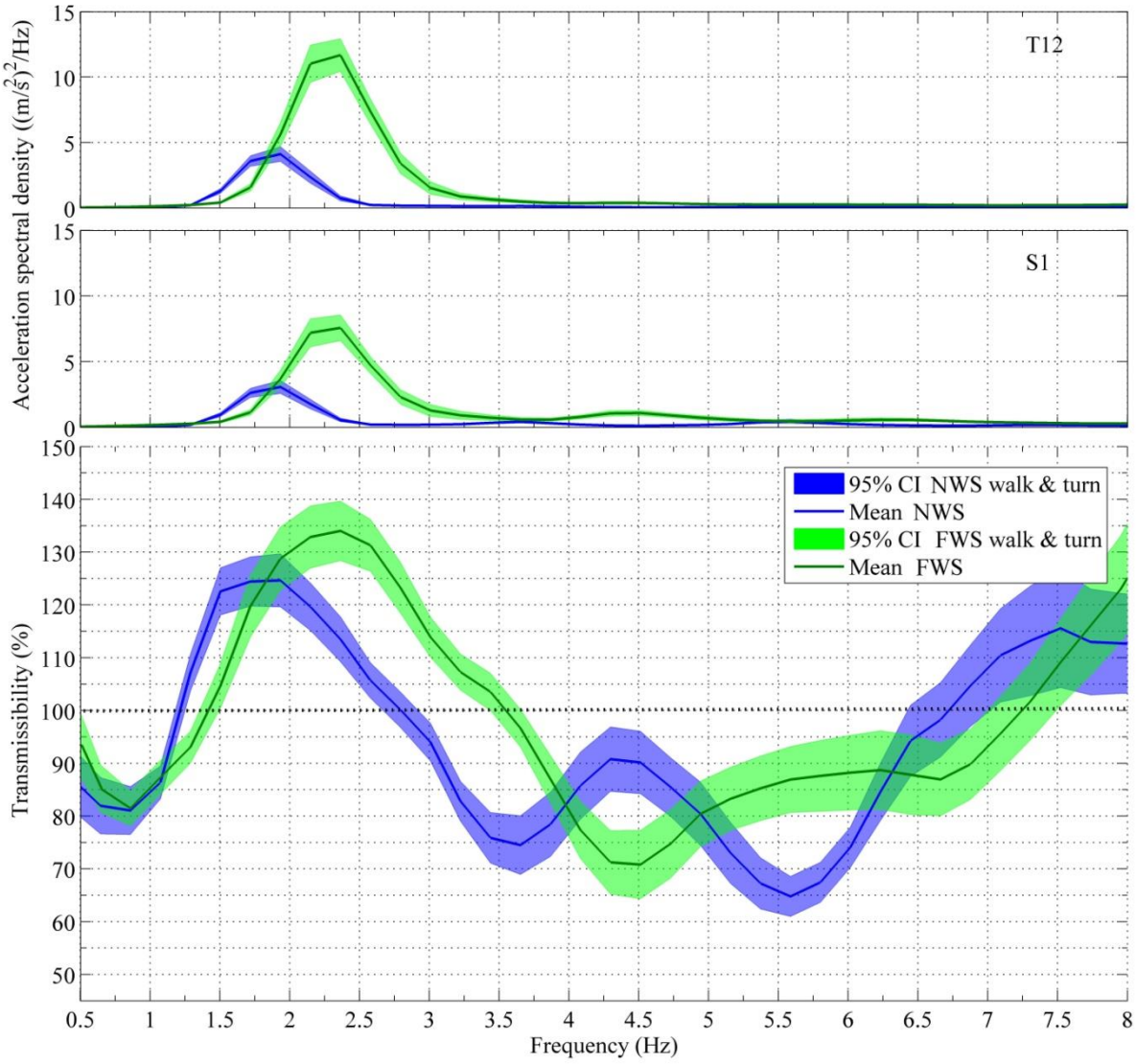

Figure 6.3-1 Input (S1) and output (T12) acceleration PSD and transmissibility during combined walking and turning. Normal walking speed (NWS) and fast walking speed (FWS) for the young and healthy group

The OO lumbar spine was affected by fast walking by significantly increasing transmissibility (between 2% and 9%) in more frequency intervals and physical activities compared with the YH and OH lumbar spines (Figure 6.3-2). The greatest change was observed during *w* for the YH and OH spines from 4 to 6 Hz where vibration was attenuated (between 24% and 31% less). The YH spine transmitted a significantly greater percentage of vibration during *d* from 4 to 8 Hz (around 20% more), during *a* from 4 to 6 Hz (around 10% more) and during *m* from 2 to 4 Hz (around 9% more). The OH spine transmitted a significantly lower percentage of vibration during *w* from 4 to 8

Hz (around 30% less), during *m* from 4 to 6 Hz (around 18% less), during *a* at all frequencies except between 4 and 6 Hz (between 17% and 20% less) and during *d* from 0.5 to 4 Hz (between 3% and 6% less). Significantly greater transmissibility was seen during *w* from 0.5 to 4 Hz for the OO spine (from 1% to 4% more), as well as during *m* from 2 to 4 Hz (around 4% more), during *a* from 4 to 6 Hz (around 4% more) and during *d* from 4 to 8 Hz (between 5% and 9% more). Regarding the frequencies at which mean maximum transmissibility was found at frequency intervals, these were significantly affected by walking speed in different ways across all physical activities and in a different ways between groups (Appendix C1).

Fast walking had a significant effect on the differences in mean maximum transmissibility at frequency intervals between groups (Figure 6.3-3). FWS created new significant differences between groups where there was none before during *w* at NWS (Figure 4.3-5). Similarly at some frequency intervals significant differences between groups were suppressed during fast walking. During *w* at FWS the OO spine transmitted significantly more vibration than the YH spine between 0.5 and 6 Hz reaching a maximum of 138% (SD 32%). Similarly transmissibility of the OH spine was significantly greater than the YH spine from 0.5 to 4 Hz. The OO spine transmitted significantly more vibration than the OH one during *w* (from 4 to 6 Hz), and during *a* (from 2 to 4 Hz). On the other hand it transmitted significantly less vibration than the OH spine during *m* (from 6 to 8 Hz).

Significant differences in the percentage of vibration transmitted at frequency intervals were observed between physical activities (Figure 6.3-3). Fast walking created significant differences in transmissibility between physical activities that were not seen during normal walking speed (Figure 4.3-5). Mean maximum transmissibility during *w* was significantly different to *a* and *d* for the OO spine from 0.5 to 6 Hz and for the OH spine from 0.5 to 4 Hz. For the YH spine notable new significant differences were seen between *w* and *m* (from 0.5 to 6 Hz) and between *m* and *a* (from 0.5 to 4 Hz). Maximum transmissibility during FWS was found during *m* for all groups with a few

exceptions during *w* and *d*. In general, the amplification tendency of the lumbar spine is preserved during fast walking and for all groups.

Walking faster significantly decreased mean maximum transmissibility at maximum acceleration PSD for the YH and OH spines during *a* (Table 6.3-2). Transmissibility during other physical activities for the YH spine was not significantly affected by walking speed. Transmissibility was greater when walking at FWS for the OH and OO spines during *m*. The OO spine also transmitted a significantly greater percentage of vibration when *w* at FWS. Regarding frequencies at which maximum transmissibility at maximum acceleration PSD was found, all were significantly greater than the frequencies at NWS for all groups and physical activities.

When increasing the walking speed, maximum transmissibility of the YH spine was significantly lower than the OO during *w* and *d*. However it was no longer significantly different between the YH and OH groups during *d*. Walking speed did not affect significantly the distribution of mean maximum transmissibility at maximum acceleration PSD during *m* and *a*.

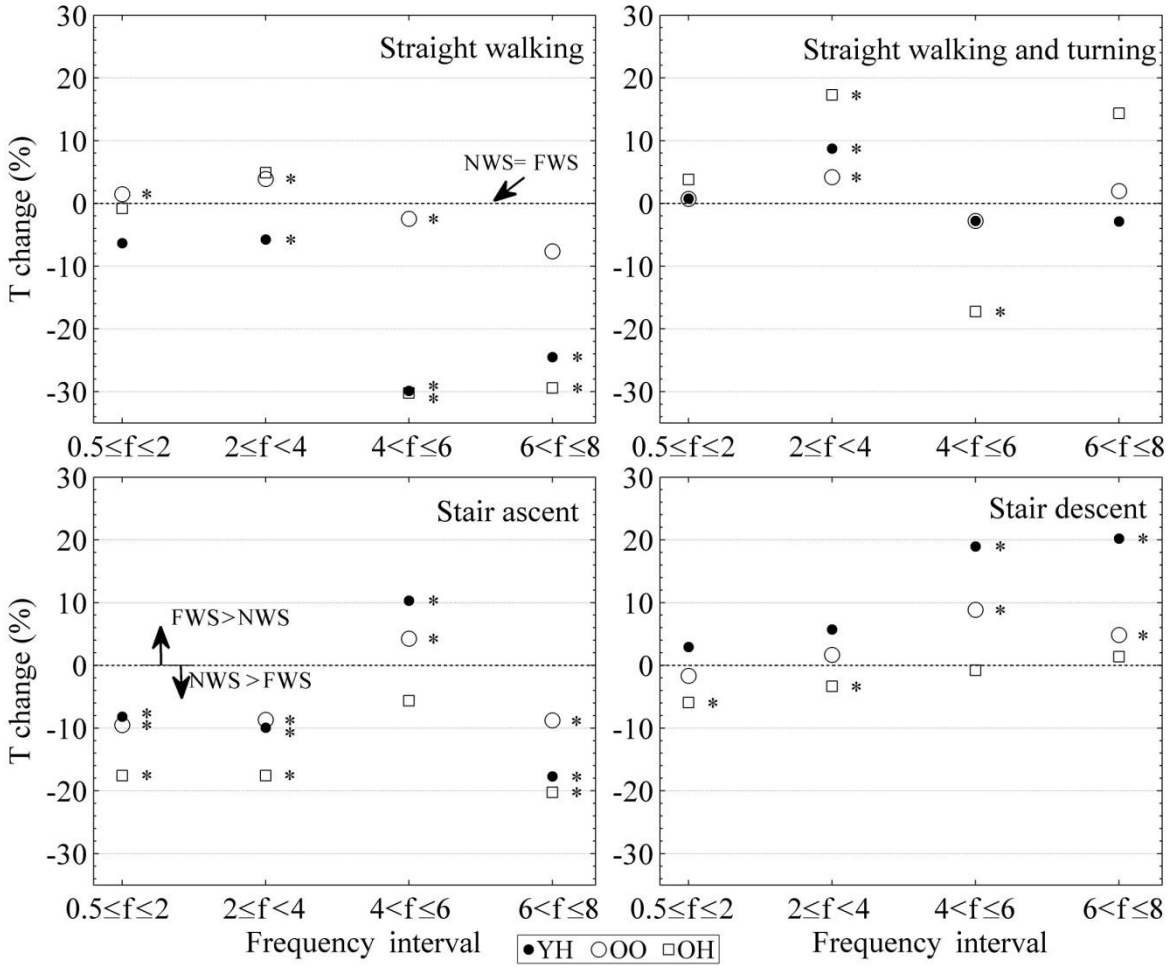

Figure 6.3-2 Relative percentage change in mean maximum transmissibility (T) for the lumbar spine when walking at fast speed. *= significant change from NWS to FWS, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

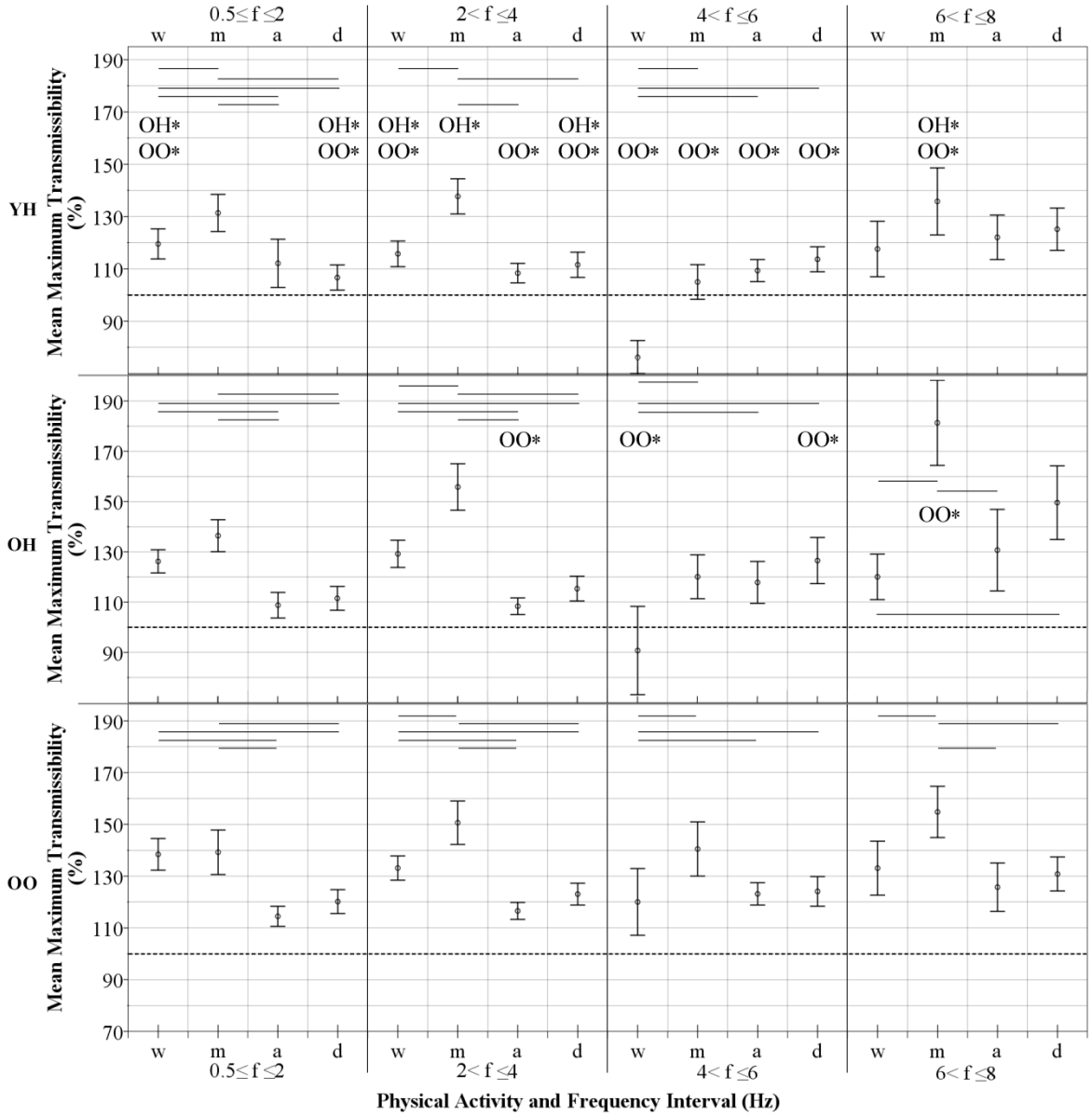

Figure 6.3-3 Mean maximum transmissibility from S1 to T12 at maximum walking speed. Effect of physical activities and osteoporosis. ─ or * = significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

Table 6.3-2 Effect of walking speed on maximum transmissibility (T) at maximum acceleration PSD for the lumbar spine

Variable	Activity	Group and walking speed								
		YH			OH			OO		
		NWS	FWS		NWS	FWS		NWS	FWS	
T (%)	w	114(21) ≈ 115(22)	=		115(14)≈ 125(19)	=		119(31) < 125(22)*		+
	m	124(23) ≈ 133(30)	=		126(14)< 149(31)*		+	129(38) < 145(49)*		+
	a	124(26) > 107(15)*		-	130(29)> 104(10)*		-	124(28)≈ 112(17)		=
	d	103(13) ≈ 112(26)	=		117(14)≈ 113(13)	=		119(17)≈ 122(21)		=
f (Hz)	w	1.99(0.12) < 2.41(0.25)*		+	1.99(0.11)< 2.47(0.51)*		+	2.04(0.17) < 2.46(0.58)*		+
	m	1.86(0.14)< 2.34(0.25)*		+	1.99(0.46)< 2.31(0.20)*		+	1.87(0.17) < 2.60(1.89)*		+
	a	2.04(0.23) < 3.30(0.50)*		+	2.06(0.27)< 3.00(0.41)*		+	2.07(0.42) < 2.95(0.54)*		+
	d	2.35(0.35) < 3.92(0.91)*		+	2.22(0.35)< 3.07(0.65)*		+	2.22(0.50) < 3.18(1.38)*		+

(-) decrease, (+) increase or (=) no significant change in frequency from normal (NWS) to fast walking speed (FWS), mean (SD), *=significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

6.3.3. Thoracic spine

An example of mean transmissibility curves and acceleration PSD ($(m/s^2)^2$/Hz) is presented with 95% confidence intervals for combined walking and turning at NWS and FWS for the YH group (Figure 6.3-4). Maximum acceleration PSD at maximum walking speed is slightly more than twice the magnitude of acceleration PSD when NWS (Figure 6.3-4). Attenuation was mainly observed above approximately 1.25 Hz. Similarly, walking at a fast speed seemed to attenuate a greater percentage of transmissibility than walking at a normal speed from approximately 0.5 to 1.2 Hz and from approximately 2 to 6.25 Hz. Maximum transmissibility was observed at low frequency (below 1 Hz). The greatest attenuations were observed at maximum acceleration PSD.

The percentage change in mean maximum transmissibility at frequency intervals from NWS (Figure 5.3-5) to FWS is presented in Figure 6.3-5. The YH spine was the one that was mainly affected by fast walking by decreasing transmissibility significantly at most physical activities and frequency intervals in comparison with the older spines (OH and OO). Fast walking significantly increased transmissibility of the YH spine only during *a* from 0.5 to 2 Hz (around 4% more). The OH spine transmitted significantly less vibration during *w* at all frequencies (from 11% to 34 % less), during *m* from 0.5 to 6 Hz (from 13% to 17%), during *a* from 2 to 8 Hz (from 8% tom 25% less) and during *d* from 6 to 8 Hz (around 12% less). The OO spine transmitted significantly less vibration during *w* from 0.5 to 6 Hz (from 16% to 35% less), during *m* from 0.5 to 4 Hz (from 12% to 15% less), during *a* from 2 to 8 Hz (around 12% less) and during *d* from 4 to 6 Hz (around 9% less). The frequencies at which mean maximum transmissibility was found at frequency intervals, were significantly affected by walking speed in different ways across all physical activities and in a different manner between groups (Appendix C2).

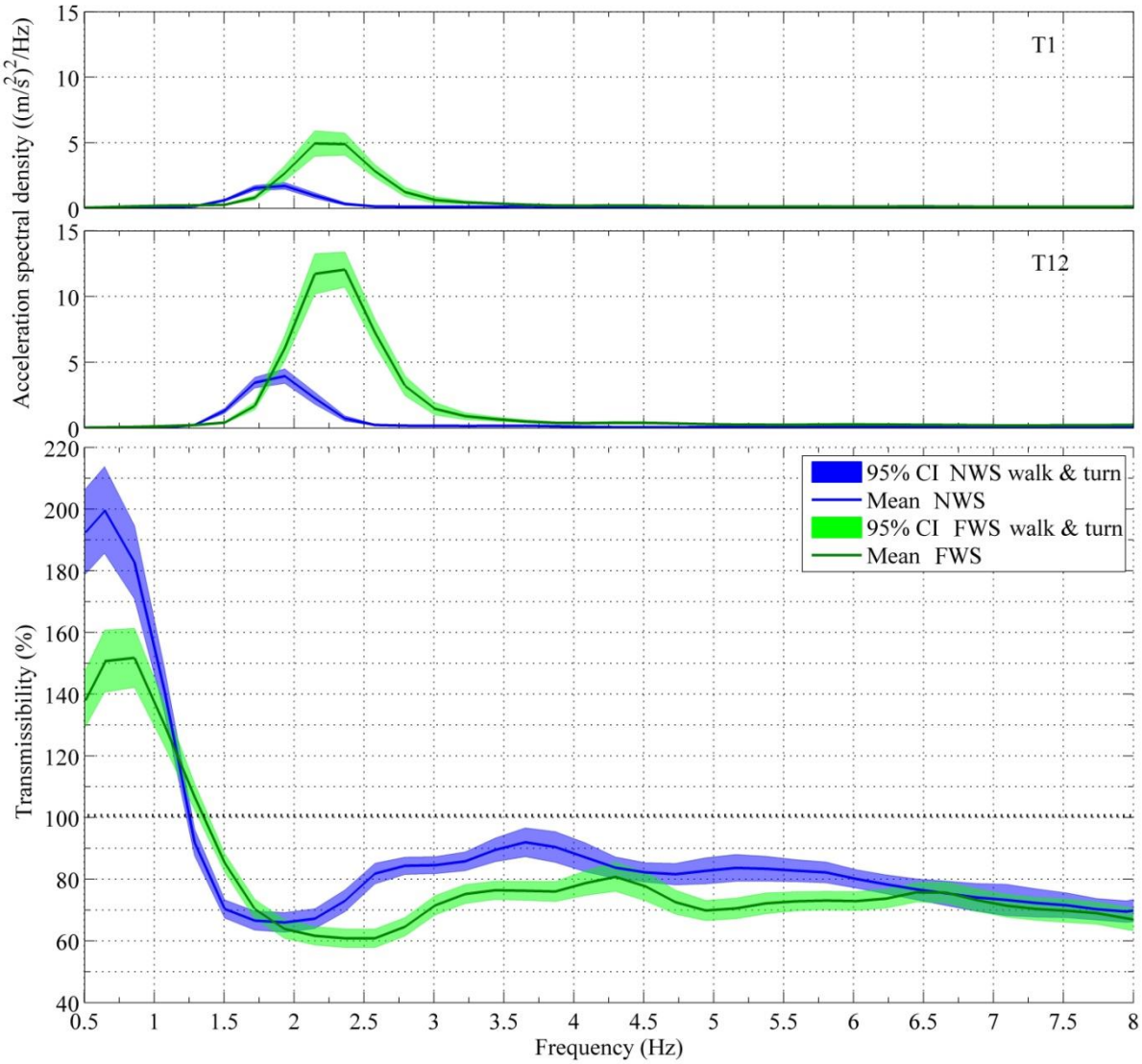

Figure 6.3-4 Input (T12) and output (T1) acceleration PSD and transmissibility during combined walking and turning. Normal walking speed (NWS) and fast walking speed (FWS) for the young and healthy (YH) group

Fast walking had a significant effect in the differences in mean maximum transmissibility at frequency intervals between groups (Figure 6.3-6). Walking at FWS created new significant differences between groups where there was none before when walking at NWS. Similarly, at some frequency intervals significant differences between groups were suppressed during FWS. Mean maximum transmissibility of the YH spine was significantly greater than the OO spine from 0.5 to 2 Hz during *a*. Similarly, it was significantly greater than the OH spine from 0.5 to 4 Hz during the same activity. Mean maximum transmissibility of the OO spine was significantly greater than transmissibility of the OH spine during *w* at FWS from 0.5 to 4 Hz. The greatest transmissibilities

observed from 0.5 to 2 Hz for the YH, OH and OO groups were 160 ± 47%, 172 ± 53% and 172 ± 53% respectively.

Fast walking had a significant effect on transmissibility differences between physical activities mainly for the OO spine (Figure 6.3-6). Mean maximum transmissibility was significantly different between *w* and *m* from 0.5 to 2 Hz for the OO spine when walking at FWS. Walking speed had no effect on the significant differences between physical activities seen during NWS (Figure 5.3-5) for the OH spine (from 2 to 4 Hz) and for the YH spine (from 2 to 6 Hz). Mean maximum transmissibility was significantly different between *w* and *m* for the OH spine (from 0.5 to 2 Hz). In general, the attenuation tendency of the thoracic spine was also observed during fast walking for all groups; however the percentage of vibration transmitted decreased significantly when compared with that measured at NWS (Figure 5.3-5).

Fast walking had no significant effect on mean maximum transmissibility at maximum acceleration PSD for the OH and OO spines during all physical activities (Table 6.3-3). On the contrary, it significantly increased transmission of vibration during *a* and decreased it significantly during *d* for the YH spine. Regarding frequencies at which mean maximum transmissibility at maximum acceleration PSD was found, all were significantly greater than the frequencies at a normal walking speed for all groups and physical activities. When increasing walking speed, maxT@maxPSD remained significantly different between the YH and OO groups during *d* and was no longer significantly different between the YH and OH group for the same physical activity. The differences between groups during *w*, *m* and *a* were not significantly affected by walking speed.

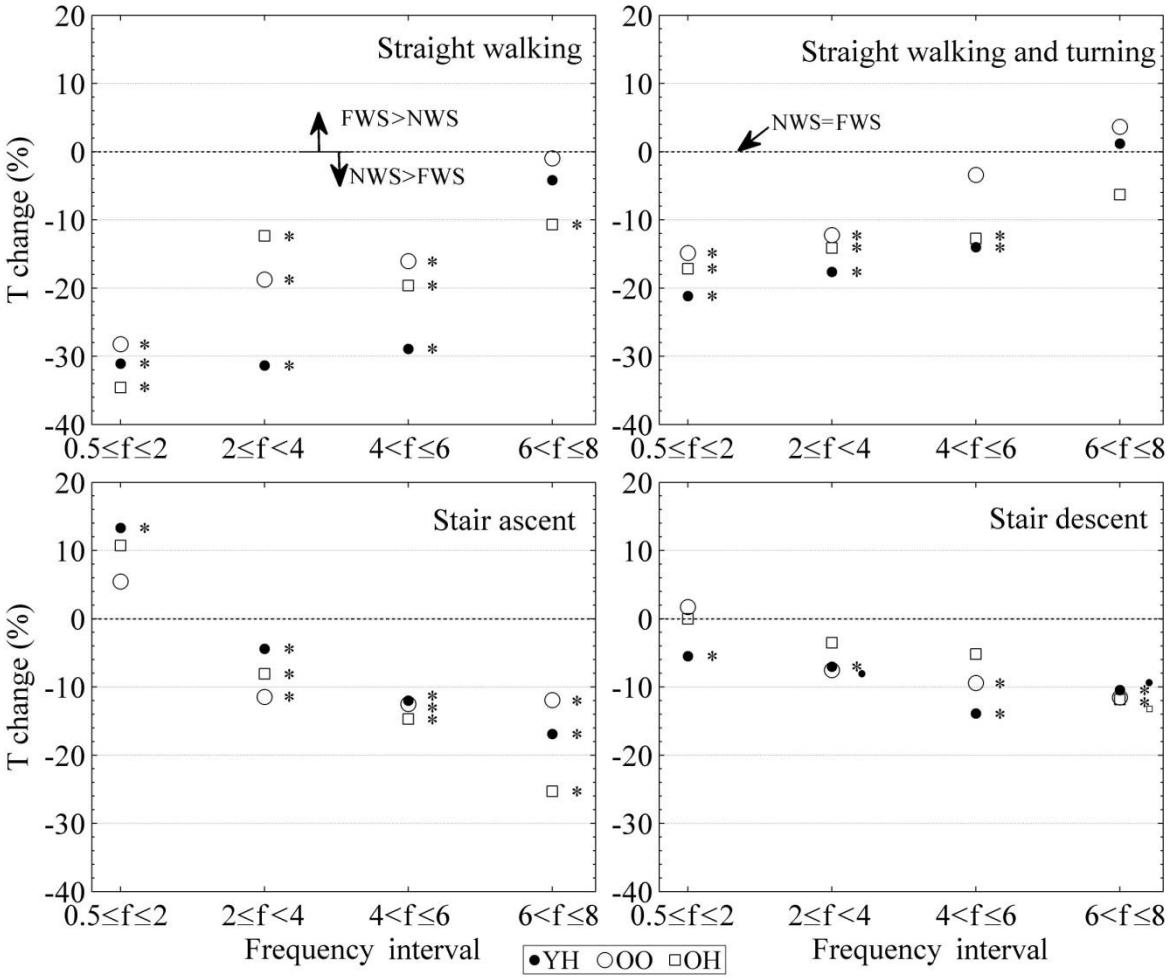

Figure 6.3-5 Relative percentage change in mean maximum transmissibility (T) for the thoracic spine when walking at fast speed. *= significant change from NWS to FWS, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

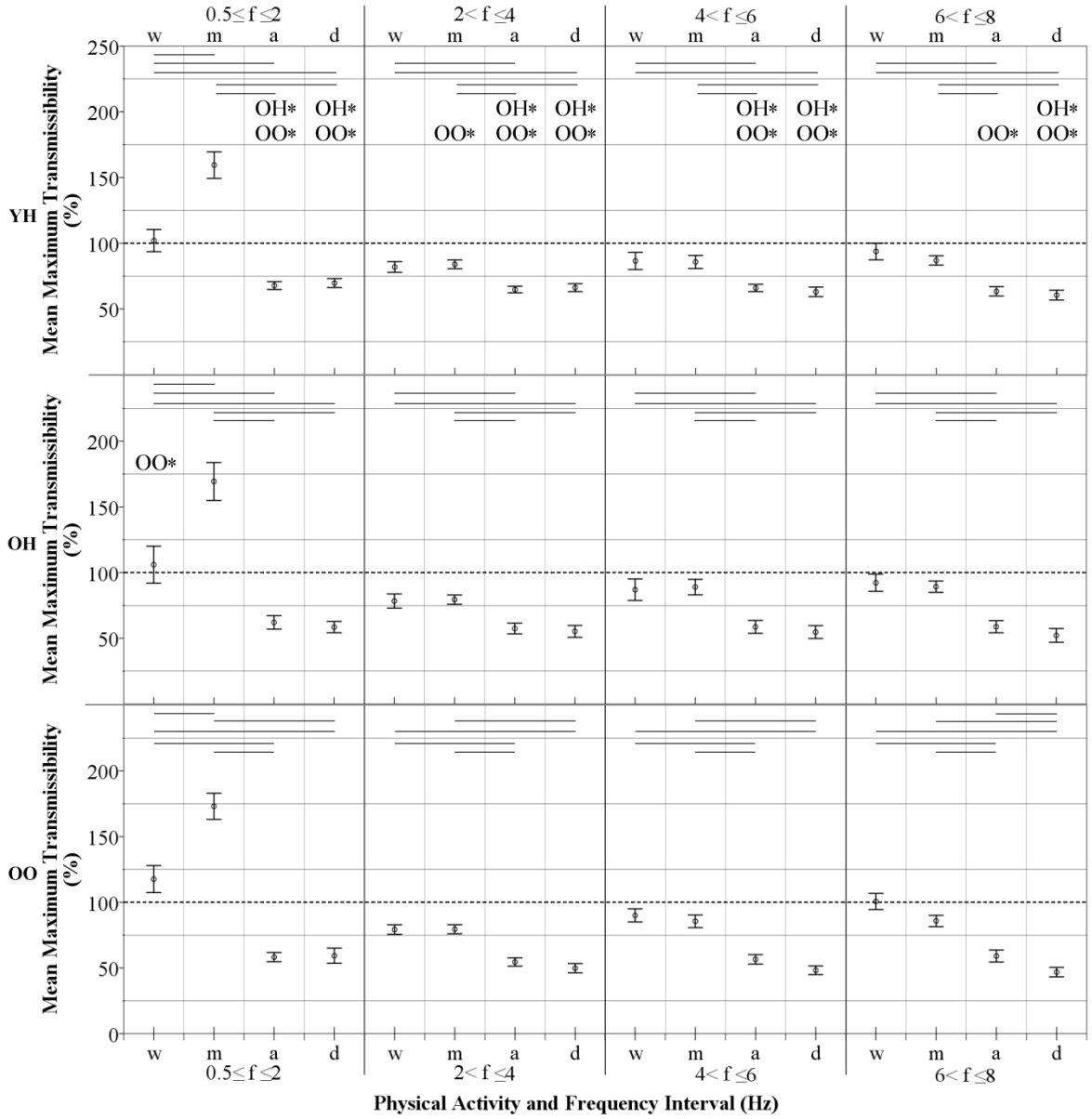

Figure 6.3-6 Mean maximum transmissibility from T12 to T1 at maximum walking speed. Effect of physical activities and osteoporosis. ─ or *= significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

Table 6.3-3 Effect of walking speed on maximum transmissibility (T) at maximum acceleration PSD for the thoracic spine

Variable	Activity	Group and walking speed								
		YH			OH			OO		
		NWS	FWS		NWS	FWS		NWS	FWS	
T (%)	w	66(12) ≈ 66(10)	=		59(12) ≈ 56(19)	=		59(15) ≈ 60(15)	=	
	m	62(11) ≈ 58(11)	=		57(13) ≈ 58(25)	=		56(15) ≈ 56(14)	=	
	a	54(10) < 61(11)*	+		53(14) ≈ 55(20)	=		55(13) ≈ 53(15)	=	
	d	67(11) > 61(14)*	-		56(16)≈ 54(21)	=		53(16) ≈ 48(17)	=	
f (Hz)	w	1.97(0.13) < 2.42 (0.23)*	+		2.06(0.26) < 2.55(0.55)*	+		2.02(0.17) < 2.46(0.33)*	+	
	m	1.83(0.13) < 2.30(0.21)*	+		2.32(0.97) < 2.57(0.86)*	+		1.87(0.19) < 2.48(1.10)*	+	
	a	1.98(0.24) < 3.49(0.67)*	+		2.30(0.61) < 3.00(0.44)*	+		2.47(0.67) < 2.99(0.48)*	+	
	d	2.31(0.35) < 4.07(1.40)*	+		2.29(0.50)< 2.94(0.69)*	+		2.18(0.47) < 2.96(0.87)*	+	

(-) decrease, (+) increase or (=) no significant change in frequency from normal (NWS) to fast walking speed (FWS), mean (SD), *=significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

6.3.4. RMS acceleration

Percentage change in mean RMSa from NWS (Figure 4.3-2 and Figure 5.3-2) to FWS for all groups, physical activities and spine locations is presented in Figure 6.3-7. Significant effect of walking speed on RMS acceleration magnitudes is indicated with an asterisk. Mean RMS acceleration was significantly increased when fast walking for the YH group (between 30% and 190% more), except for *w* at T8 and for *d* at S1. For the OH group, mean RMSa was significantly increased when walking fast (between 45% and 170% more), except during *w* and *d* at T8. Interestingly, a significant decrement (around 10%) was seen during *w* at T8 for the OH group. Mean RMS acceleration was significantly increased when fast walking (between 50% and 135% more) during all physical activities and at most spinal levels for the OO group.

Fast walking had a significant effect on the differences in RMSa between physical activities and between groups Figure 6.3-8. Median RMS acceleration is presented for all groups, physical activities and spine locations. FWS created new significant differences between groups where there was none when walking at NWS (Figure 4.3-2 and Figure 5.3-2). The YH spine received greater

RMSa compared with the OO spine during *w* and *a* at T12. The YH spine received greater RMSa at S1 during *w* and *m* in comparison with the OH spine. RMSa at T12 was no longer different between the YH and OH spines during *a* and at S1 during *d*. Walking fast also cleared previous significant differences between groups found at NWS (Figure 4.3-2 and Figure 5.3-2). The YH spine transmitted the same magnitude of RMSa as the OO spine during *w* at S1 and *d* (T8, T12 and S1). The OH spine no longer received significantly greater RMSa than the OO spine during *m* (T8 and T12). Similarly the OO spine no longer received significantly greater RMSa than the OH spine during *a* at S1. Walking speed had no significant effect on the differences in RMSa between groups for all physical activities for T1. Similarly it had no effect during *w* and *a* for T8. The greatest RMSa achieved during FWS for all groups, all locations of the spine and physical activities was during *a*, except for the OO spine at T8 where this was greater during *d*. These greatest RMSa magnitudes were previously observed during *w* and *d* while walking at NWS (Figure 4.3-2 and Figure 5.3-2).

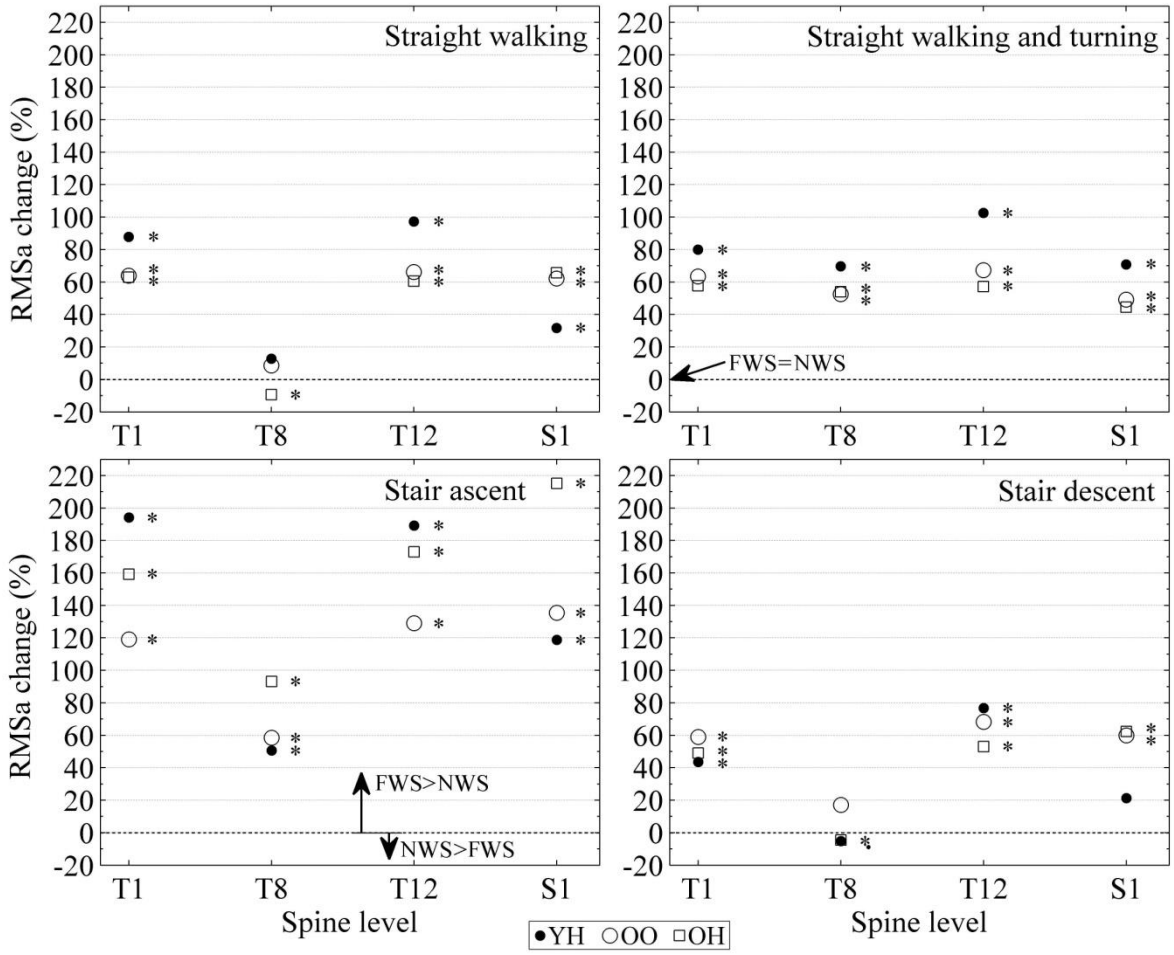

Figure 6.3-7 Relative percentage change in RMSa from NWS to FWS. *= significant change, young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), eight thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1)

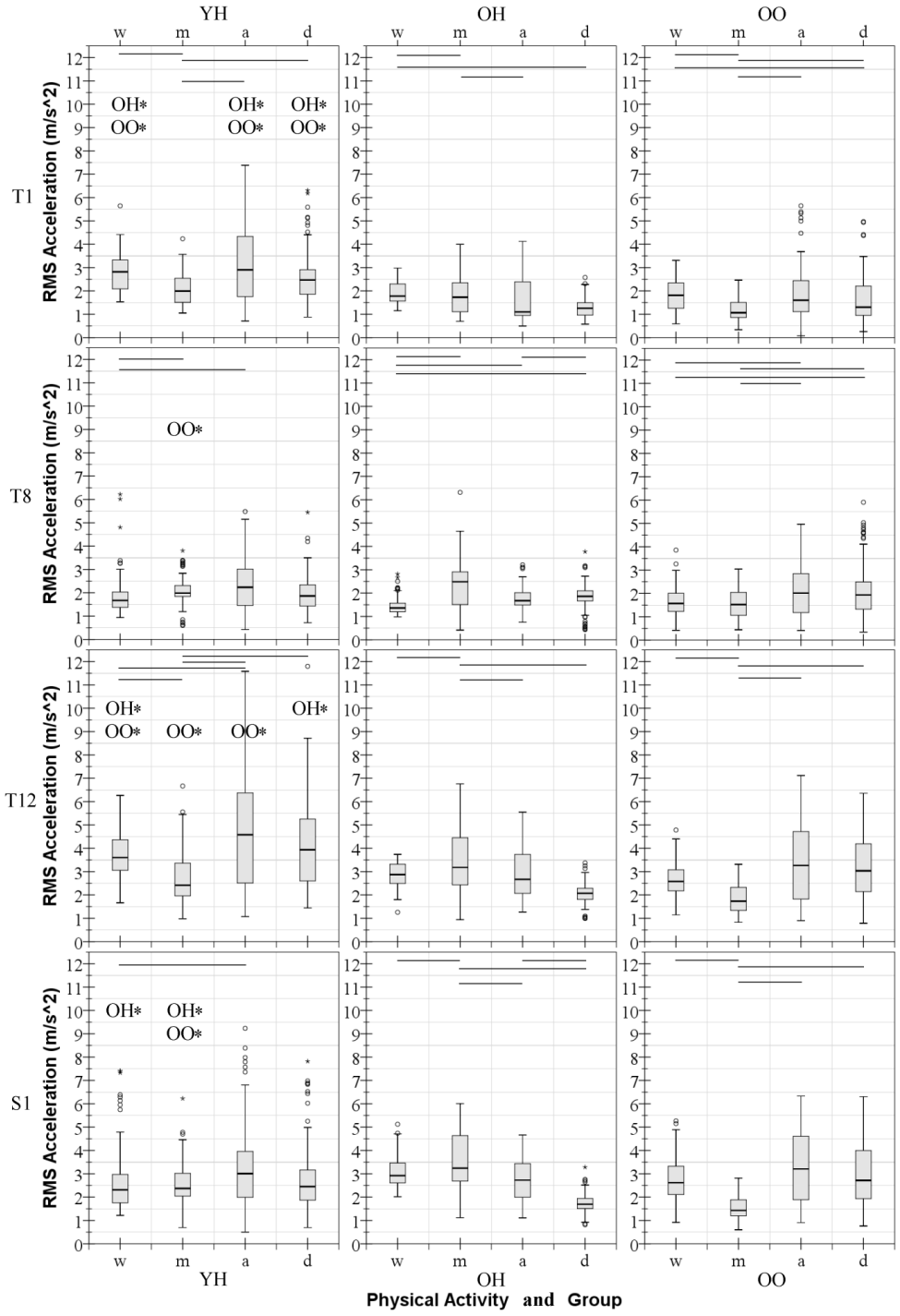

Figure 6.3-8 Root mean square acceleration at maximum walking speed. Effect of physical activities and osteoporosis. ─ or *= significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), eight thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1), straight walking (w), walking and turning (m), stair ascent (a), stair descent (d)

6.3.5. Differences between lumbar and thoracic spines

Changes in the significant differences in mean maximum transmissibility at frequency intervals between lumbar and thoracic spine were observed during *w* at FWS (Figure 6.3-9). Only changes in differences in transmissibility between lumbar and thoracic spines due to fast walking are mentioned here, differences that were not affected remain as reported previously. The YH thoracic spine was previously transmitting the same percentage of vibration as the thoracic section during *w* at NWS. When walking at FWS, it transmitted significantly less (≈14%) than the lumbar (from 0.5 to 2 Hz) and significantly more (≈16%) than the lumbar from 4 to 6 Hz. The OH thoracic spine was no longer transmitting more than the lumbar (from 0.5 to 2 Hz) and less from 4 to 6 Hz but significantly less (≈16%) and equally as the lumbar spine during *w*. Walking speed significantly affected the transmissibility differences between the OO lumbar and thoracic spines from 0.5 to 2 Hz, where the thoracic transmitted significantly less vibration (≈16%). During *w*, the YH thoracic spine transmitted significantly less vibration (≈16%) than the lumbar spine from 4 to 6 Hz. During *d*, all thoracic spines (YH, OH and OO) transmitted significantly less vibration than the lumbar sections, from 36% (at NWS) to 64% due to FWS. During *a*, the YH thoracic spine transmitted even less vibration than the lumbar section due to FWS, around 38% compared with 24% during NWS.

Walking speed had no effect on the significant differences in mean maximum transmissibility at maximum acceleration PSD between lumbar and thoracic spines; the tendency of the thoracic spine of transmitting less vibration than the lumbar spine across physical activities and groups was preserved (Appendix C4).

Regarding the differences between lumbar and thoracic spines for frequencies at which mean maximum transmissibility was found, these were significantly affected by walking speed (Figure 6.3-10). Only changes in differences in frequencies at which transmissibility was found between lumbar and thoracic spines due to fast walking are mentioned here, differences that were not affected remain as reported previously. The YH thoracic spine transmitted at significantly lower frequencies than the lumbar spine during *w* from 4 to 6 Hz (≈3% lower), during *a* from 2 to 6 Hz

(from 3% to 8% less) and during *d* from 2 to 6 Hz (from 12% to 22% less). On the other hand the YH thoracic spine transmitted vibration at a significantly higher frequency from 4 to 6 Hz during *m* when compared with the lumbar spine (≈1% higher). The OH thoracic spine no longer transmitted vibration at significantly lower frequency compared with the lumbar during *w* (from 6 to 8 Hz) and *a* (at all frequencies except from 2 to 4 Hz), it transmitted vibration at the same frequency as the lumbar spine instead. The OH thoracic spine transmitted vibration at significantly higher frequencies (≈16% higher) between 2 and 4 Hz during *a*. On the other hand it transmitted vibration at a significantly lower frequency (≈22% lower) in the same frequency interval but during *d*. The OO thoracic spine transmitted vibration at the same frequency as the lumbar during *w* and during *a* from 6 to 8 Hz. It also transmitted vibration at a significantly lower frequency during *a* from 2 to 4 Hz (≈22% lower) and during *d* from 2 to 6 Hz (from 4% to 8% lower).

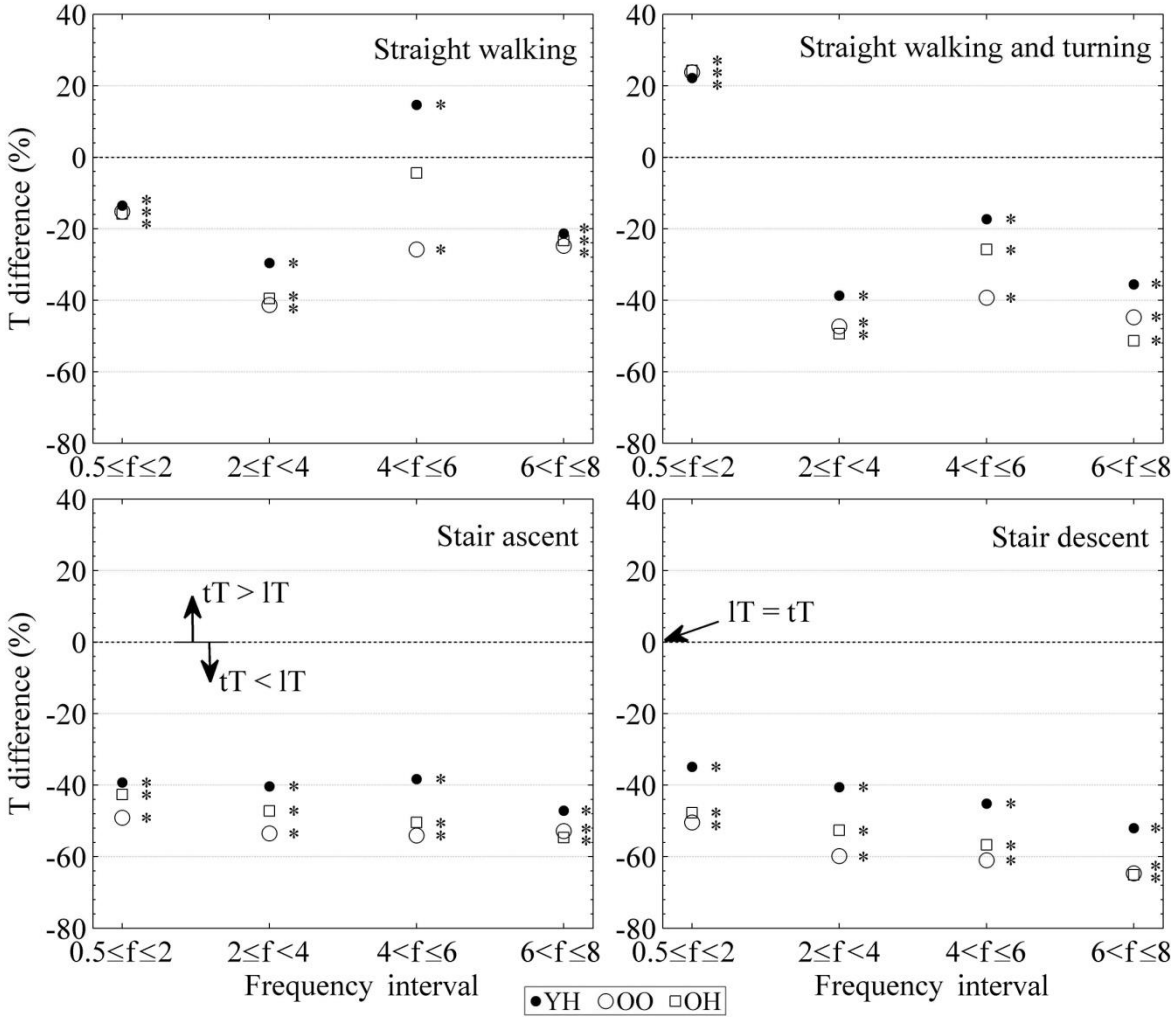

Figure 6.3-9 Percentage difference in mean maximum transmissibility at frequency intervals between the thoracic spine (tT) and the lumbar spine (lT). *=significant difference in T between lumbar and thoracic spine.FWS, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

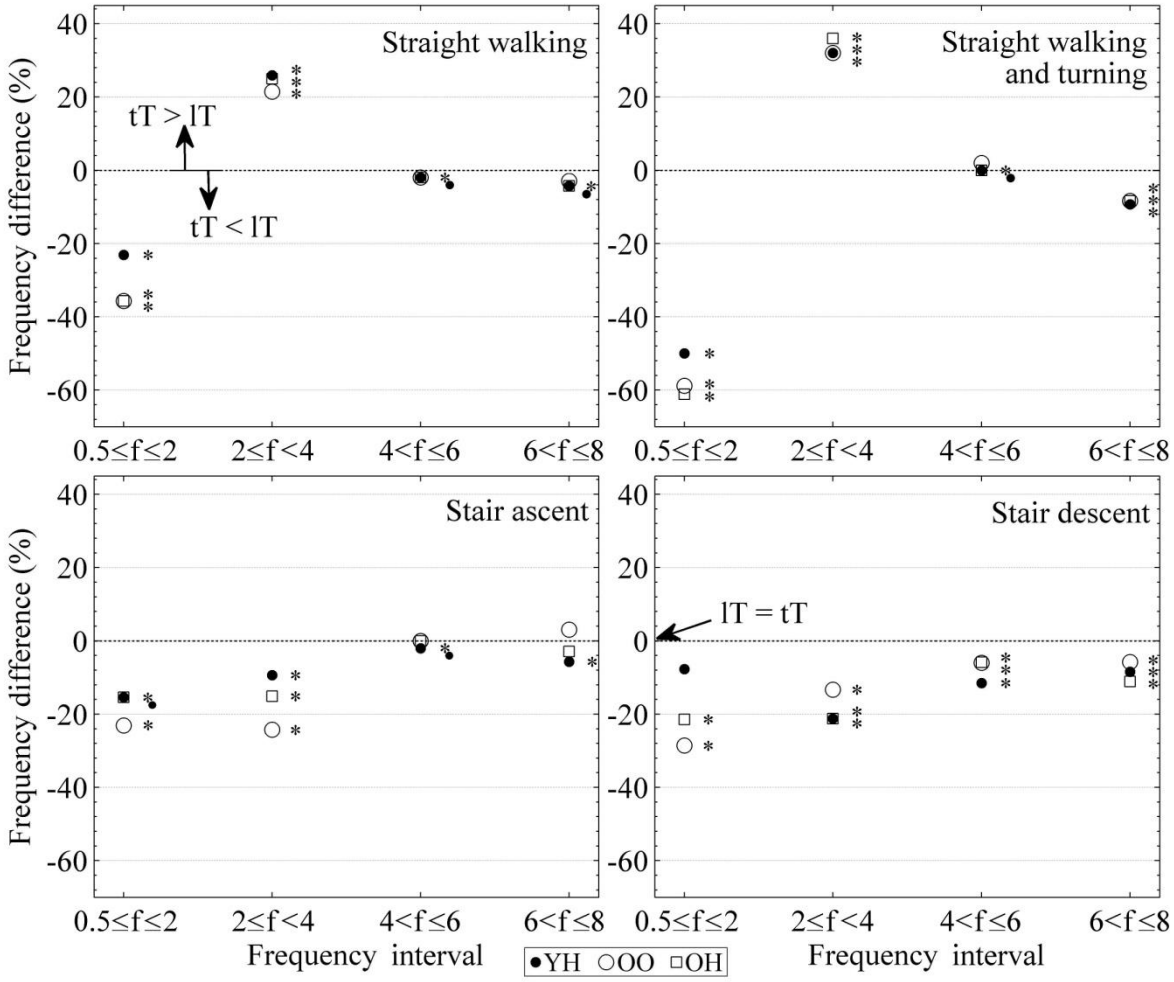

Figure 6.3-10 Percentage difference in the frequency (f) at which mean maximum vibration transmissibility at frequency intervals was found between the thoracic spine (tf) and the lumbar spine (lf). *=significant difference in f between lumbar and thoracic spine, fast walking, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

6.3.6. Transmissibility Predictors

Prediction of maximum transmissibility at maximum acceleration PSD dependant of RMSa, spinal curvatures, BMD, age, gender and BMI was performed for lumbar and thoracic spines. Gender was considered as a predictor for maximum transmissibility at maximum acceleration PSD only for the young and healthy spine only. Forward multiple regressions were performed for transmissibility obtained at self selected fast walking speed (Table 6.3-4). Based on the value of the correlation coefficient, RMSa at S1 was the most important predictor for the lumbar spine of all groups. RMSa at T1 was the most important predictor for the transmissibility of the thoracic spine of all groups.

The percentage of vibration transmitted that was predicted during NWS was decreased with fast walking for the YH and OH groups and for both lumbar and thoracic spines (Figure 4.3-5 and Figure 5.3-2). The percentage of transmissibility variability explained for the OO group increased from 15.3% to 17.3% for the lumbar spine and from 64% to 66.8% for the thoracic spine. The predictors involved in the models changed from NWS (Table 4.3-3 and Table 5.3-2) to FWS (Table 6.3-4). Lumbar lordosis became an important and significant contributor of transmissibility during fast walking for the thoracic spine of the YH and OH groups. This was not the case when walking at a normal self selected walking speed (Table 4.3-3). Specific transmissibility explained by the spinal curves increased with fast walking speed except for the OH lumbar spine for which the lumbar lordosis had neither a significant nor important contribution to the model. Notable increments in transmissibility variability explained by spinal curvatures are for the thoracic spines which presented percentages between 5.9% and 10.7%.

Table 6.3-4 Predictors of transmissibility during fast walking for young and older spines

Spine	Group	r^2	Residual	Variability explained	Model Term	Coefficient	Significance	Importance
Lumbar	YH	0.237	6.292	23.7%	Intercept	1.110	0.000	
					Female	0.144	0.001	0.207
					RMSa at S1	-0.046	0.002	0.205
				3.46%	Lumbar Lordosis	0.004	0.015	0.199
					RMSa atT12	-0.021	0.058	0.195
					WS	0.047	0.106	0.194
	OH	0.401	3.246	40.1%	Intercept	1.484	0.000	
					WS	0.180	0.000	0.235
					RMSa at S1	-0.169	0.000	0.222
					RMSa at T12	0.082	0.015	0.185
					Age	-0.005	0.089	0.179
					T-score	-0.073	0.098	0.179
	OO	0.173	13.217	17.3%	Intercept	1.042	0.001	
					BMI	0.032	0.000	0.257
					RMSa at S1	-0.069	0.000	0.256
					Age	-0.008	0.015	0.244
					WS	0.084	0.025	0.243
Thoracic	YH	0.390	1.177	39%	Intercept	0.875	0.000	
					RMSa at T12	-0.038	0.000	0.139
					RMSa at T1	0.070	0.000	0.138
				9.8%	Thoracic Kyphosis	0.005	0.000	0.131
				6.3%	Lumbar Lordosis	0.003	0.000	0.125
					RMSa at S1	-0.002	0.006	0.120
					RMSa at T8	0.027	0.052	0.116
					BMI	-0.006	0.101	0.116
					T-score	0.021	0.127	0.115
	OH	0.287	1.777	28%	Intercept	0.346	0.027	
					RMSa at T1	0.140	0.000	0.216
					RMSa at T12	-0.078	0.000	0.203
				10.7%	Thoracic Kyphosis	0.005	0.000	0.202
				5.9%	Lumbar Lordosis	0.004	0.007	0.191
					Age	0.005	0.011	0.189
	OO	0.668	1.461	66.8%	Intercept	0.739	0.000	
					RMSa at T1	0.183	0.000	0.335
					RMSa at T12	-0.102	0.000	0.280
				9.22%	Thoracic Kyphosis	0.004	0.000	0.211
					T-score	0.020	0.036	0.174

Root mean square acceleration (RMSa), walking speed (WS) young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), eight thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1)

6.4. Discussion

The purpose of this study was to determine if fast walking had a significant effect on the percentage and magnitude of vibration transmitted by the lumbar and thoracic spines during physical activity for people with different BMD. Mean maximum transmissibility examined at specific frequency intervals has provided evidence that fast walking has a significant effect on the mechanical response of the spine during physical activities. In general, the lumbar spine amplified the percentage of vibration transmitted while the thoracic spine attenuated it for all groups.

Notable effects of fast walking on the lumbar spine are: (1) a greater percentage of vibration is transmitted by the OO lumbar spine in comparison with normal walking speed, (2) vibration is attenuated during straight walking from 4 to 6 Hz by the YH and OH spines and (3) an increase in the differences between physical activities when compared with normal walking speed. It is suggested that osteoporosis is associated with an increased percentage and magnitude of vibration transmitted during fast walking compared with normal walking speed at the lumbar spine. Important effects of fast walking on the response of the thoracic spine are: (1) significant decrease in transmission of vibration and (2) decreased vibration attenuation by the OO and OH thoracic spines in comparison with a young and healthy one. Attenuation of vibration is possible by increasing walking speed (at specific frequencies and physical activities). Therefore a thorough characterization of the mechanical response of the individual osteopenic and osteoporotic spines is highly recommended.

When increasing walking speed, the OO and OH lumbar spine tended to transmit a greater percentage of vibration than the YH. The YH lumbar and thoracic spines reduced vibration transmissibility during fast walking. The OO thoracic spine did not reduce vibration transmissibility. Attenuation capacity was reduced with ageing at lumbar and thoracic spines regardless of osteoporosis. It is suggested that the effect of collagen network deterioration experienced due to age is greater than the effects of osteoporosis (Yerramshetty and Akkus, 2008, Orkoula et al., 2012).

Fast walking increased the magnitude of the vibration transmitted to most locations of the lumbar and thoracic spines (Figure 6.3-7). Yet the OH spine at the level of T8 experienced a reduction in vibration magnitude during fast walking in a straight line. This may have implications on the incidence of vertebral fractures at this level of the spine (Ravishankar, 2009). Other clinically relevant observations involve the OO and OH spines which transmitted significantly less vibration magnitude when compared with the YH spine (Figure 6.3-8). It is imperative to determine if this reduced vibration magnitude is a factor contributing to weaker bone (therefore increasing the risk of fractures) or if it is enough to maintain the necessary bone. Current evidence suggesting that exercise induces a small increase in spinal BMD supports the observation in this study that the diminished magnitude of vibration measured is indeed insufficient to maintain or significantly increase spinal BMD (Hamilton et al., 2010a, IOF, 2012, Gómez-Cabello et al., 2012, Cheung and Giangregorio, 2012). This effect will have to be further investigated through a bigger sample of the population and supported by medical imaging techniques to explain this response in terms of volumetric material and structural properties before being considered as a factor during the prescription of physical activity. Unfortunately these results cannot be compared with previous data since this is the first time RMSa is measured in the older population with and without osteoporosis during fast walking.

The clinical implications of older spines receiving (1) more, (2) less or (3) the same percentage of vibration as a young and healthy spine during fast walking have to be understood before being able to attempt the recommendation of individual exercise for the treatment of osteoporosis. Given that the older spines experience the same vibration transmission as a young and healthy spine, it is necessary to ask if that stimulation is neutral, beneficial or harmful for bone. The same applies for greater or lower stimulation. Before this study it was not possible to tell with certainty the effect of fast walking on vibration transmission of the spine. As shown in this study, there are physical activities and frequencies at which fast walking will not be enhancing the transmission of vibration but diminishing it. This improves our knowledge towards the formulation of optimal exercise programmes for people with spinal osteopenia and osteoporosis.

6.4.1. Differences between lumbar and thoracic spines

Mean maximum transmissibility at frequency intervals indicated significant differences between lumbar and thoracic spines during fast walking speed. This indicates that fast walking does not influence the already established response of lumbar and thoracic sections (amplification and attenuation respectively). The geometric dimensions of lumbar vertebrae increase through adult life and decrease at the thoracic spine of postmenopausal women (Kolta et al., 2012). This may contribute to the differences in transmissibility seen between the lumbar and thoracic spines. The combination of different vertebral sizes (according to sex and spine level) and the fact that osteoporosis affects first the surface of vertebral bone (Bouxsein and Genant, 2012) may lead to a varied vibration transmission response between individuals. Thus the biomechanical response of the spine will be unique for each individual. Significant differences in transmissibility between lumbar and thoracic spines were expected due to the knowledge that vertebral fractures occur most often at the thoracic spine (T5-T9, T7-T8) and at the junction of thoracic with lumbar spine (L5-T12) (Ravishankar, 2009, Waterloo et al., 2012). The YH thoracic spine transmits less vibration than the lumbar spine during straight walking from 0.5 to 2 Hz. This same pattern is repeated in the older spines. It would be arguable to say that this same response during fast walking might not be desirable for the older spines if our interest was to stimulate the spine through this physical activity. When clinicians recommend 'brisk' or fast walking (Daniels, 2008, Martyn-St James and Carroll, 2009, Langsetmo et al., 2012, UK Chief Medical Officers, 2012, Winter-Stone, 2005, WHO, 2012, NOF, 2010, NOF, 2012, NOS, 2012, Cheung and Giangregorio, 2012, Van Norman, 2010) in order to maintain BMD of the older spine, it is not clear whether this response of the thoracic spine is protecting the older (potentially weak) spine from harmful vibration or simply preventing it from receiving the needed stimulation that would enhance bone remodelling.

6.4.2. Transmissibility predictors

Multiple regressions were employed to identify important and significant contributors of maximum transmissibility at maximum acceleration PSD produced during fast walking. Root mean square acceleration, spine curvatures, BMI, T-score, gender and walking speed proved to be important and

significant variables that could explain from 17.3%% up to 66.8% of vibration transmitted through the human spine during physical activity. Fast walking was linked with varied effects on the percentage of transmissibility predicted (increment, reduction and no effect). However, a different combination of predictors was observed during fast walking speed in comparison with to those involved during normal walking speed. Walking speed is an important predictor for the lumbar spine mainly. It is very likely that the remaining maximum transmissibility at maximum acceleration PSD variability that was not possible to explain is related to geometric and material properties of the spine that are not possible to measure in vivo during daily life physical activities. Similarly, categorical predictors may help increase the prediction of transmissibility (risk of fracture, ethnicity, alcohol and tobacco consumption, family history of osteoporosis and use of pharmacological treatments). An individual assessment of transmissibility may be more appropriate. It was not possible to use gender as a predictor during the multiple regression analysis for the older spines (healthy and osteoporotic) since more female subjects participated in this study than male subjects. Similarly, alcohol consumption and smoking habits could not be considered since most volunteers neither consumed alcohol nor smoked.

Spinal curvatures have an important contribution to the prediction of transmissibility at maximum acceleration PSD during fast walking. Along with thoracic kyphosis, lumbar lordosis becomes an important and significant predictor of transmissibility of the thoracic spine of healthy individuals. Fast walking may be associated to changes in compressive and shear forces exerted on the lumbar spine due to muscular function. This consequently may affect vibration transmitted to the thoracic spine and its dependence on lumbar lordosis. It is not possible to compare transmissibility variability with other studies, since this is the first time spinal vibration transmission is measured during fast walking.

6.4.3. Limitations

It is important to note that there is not a linear relationship between walking speed and vibration transmissibility. Spinal vibration transmission is dependent on other variables such as spinal curvature, T-score, frequency, BMI, age and vibration magnitude (Morgado Ramírez et al., 2013a,

Morgado Ramírez et al., 2013b). Added to this, there is high variability in vibration transmission between subjects. Previously reported self selected and normal speed is below what was measured, potentially due to subjects maintaining a healthy life style through lifespan (Boyer et al., 2012). Boyer et al. (2012) also reported maximum walking speeds been 1.88 m/s (young) and 1.85 m/s (older), been below the maximum walking speed reported here. Only the self selected maximum walking speed of older adults was within speeds reported here for the older osteoporotic group. Individuals participating in this study may have maintained a healthy life style through their lifespan, thus future research may benefit from seeking the participation of frailer individuals.

6.5. Conclusion

The older lumbar spine may receive greater stimulation than a young and healthy one during fast walking. The older thoracic spine may receive lower stimulation during fast walking. These are reasons to consider the thorough re-evaluation of the clinical effect of that vibration delivered to a spine that is experiencing degeneration due to age and osteoporosis. During fast walking the lumbar spine amplifies vibration transmitted while the thoracic spine attenuates it. Vibration could be controlled in order to either prevent fractures during exercise or promote stronger bones through physical activity prescribed individually.

Increasing walking speed neither modifies the vibration amplification tendency of the lumbar spine nor the attenuation tendency of the thoracic spine. Root mean square acceleration, spine curvatures, BMI, T-score, gender and walking speed are important and significant variables that explain a maximum 66.8% of vibration transmitted through the human spine during daily life physical activities.

6.6. Key Findings

1. Fast walking both attenuates and amplifies the transmission of vibration and this effect is frequency dependant.
2. The vibration transmission measuring technique employed is able to detect changes due to walking speed.
3. During fast walking, lumbar lordosis and thoracic kyphosis remain as important predictors of vibration transmission.

CHAPTER 7 General Discussion and Conclusion

7.1. Introduction

Osteoporosis affects the lives of a considerable number of older adults making it a major concern for public health. Although physical activity is generally believed to increase or maintain BMD, its effectiveness has not been demonstrated for the spine. This is likely due to the lack of knowledge on how vibration produced during physical activity is transmitted through the body. For instance, Hill et al. (2009) pointed out that current research on vibration, as a stimulus of the human spine, lack a detailed factorial exploration of frequency and amplitude of the signals.

The present study analysed the vibration signals that are transmitted through the lumbar and thoracic spines while performing physical activity (turning, walking and stairs negotiation). Young and older male and female adults (healthy and with osteopenia and osteoporosis) volunteered for this study. Corrected data (sensor inclination and skin movement) from inertial sensors (consisting of accelerometers and gyroscopes) was used to measure vibration transmitted through the human spine during gait. Stimulation delivered to the spine during physical activity was significantly affected by ageing, osteoporosis and walking speed. The interpretation, application and limitations of the vibration measured in terms of magnitude (RMSa), transmissibility and frequency will be discussed below.

7.2. Feasibility of employing the vibration measurement technique

The pilot study performed (Chapter 3) (Morgado Ramírez et al., 2013c) indicated that skin-mounted inertial sensors attached to the spine were successfully able to measure the transmission of signals through the spine. Inertial sensors offer an economic alternative to the employment of medical imaging techniques and an ethical alternative to pins surgically inserted to bone (Rubin et al., 2003, Ziegert and Lewis, 1978, Kim et al., 1993). Previous studies (Smeathers, 1989a, Kitazaki and Griffin, 1995) had not clearly demonstrated the need for the sensor inclination correction and this study demonstrated that, after correction for skin movement and sensor inclination, the signals

were significantly different during some physical activities to those with no correction. Given the individuality of movement patterns, which vary due to physical capacity (Crosbie et al., 1997) and that spinal curvatures may significantly change with age (Singh et al., 2010), it is recommended that full correction should be used at all times.

Inertial sensors attached to the skin overlying the spinous processes of vertebrae have enabled the measurement of vibration transmission through the spine during physical activity. By providing evidence on how different physical activities produce different mechanical stimuli this study contributes to the understanding of the complex mechanical response of the spine to gait. This was further supported by studies of the lumbar and thoracic spines which employed a greater sample of the population (Chapters 4 and 5) (Morgado Ramírez et al., 2013a, Morgado Ramírez et al., 2013b). This vibration transmission measurement technique successfully identified changes due to different physical activities. It has been pointed out that studies evaluating the effects of physical activity on vertebral fractures suffer from the lack of a validated tool to adequately determine the quality or intensity of exercise for the intervention (Kemmler et al., 2013). This limitation has been tackled in this study by providing evidence of a method which can provide data on the effect of physical activity through a measurement that reflects the mechanical properties of the spine. These mechanical properties should be correlated with vertebral fractures in future fracture prevention exercise protocols.

Vibration transmissibility has been measured only in straight walking and running (Smeathers, 1989a). This study has extended the application of the vibration transmissibility measurement technique to other physical activities as stair negotiation and turning. This indicates that the vibration transmissibility measurement technique may be used for daily monitoring of physical activity. However, the current location and size of inertial sensors may interfere with daily activities such as sitting down and lying down. This limitation can be solved by optimizing the measurement of transmissibility. For example, only two sensors could be attached (at the sacrum and at the level of the first thoracic vertebra) thus reducing the number of sensors. In addition, data

acquisition could be limited to periods in which high intensity physical activity is performed thus excluding periods of inactivity.

A key reason for evaluating the feasibility of wireless inertial sensors to measure vibration transmission through the spine was that this technique offers an innovative additional application of inertial sensors for health monitoring. With the continued miniaturization of inertial sensors, it is expected that this non-invasive technique can be employed widely among the population for health monitoring purposes as well as in industry for safety monitoring purposes.

7.3. Vibration transmission

Transmissibility is the ratio of the transmitted and exciting force components in the time and frequency spectrums. Any system subjected to vibration is subjected to damping to some degree as energy is dissipated by friction and other resistances (Thomson, 1993). In that sense, transmissibility can be thought of as a measure of the stiffness of a system exposed to vibration.

It is known that mechanical damage can occur under large vibration magnitudes, for example bone fractures due to shocks, brain injury, organ haemorrhage and tearing or crushing of tissues. Other physiological effects appear if exposed to moderate vibration magnitudes, such as disorders affecting the hands and chronic injuries at the spine (Brammer and Peterson, 2003). This study has provided a quantitative analysis of the frequency components and various magnitudes that are produced and transmitted through the human spine during walking.

This study examined the previous suggestion that it is possible to find a relationship between vibration transmission and pathological changes of the human spine (Helliwell et al., 1989) and specifically, the changes due to ageing and osteoporosis were explored. The findings of this study have, in particular highlighted that physical activities produce vibration that is amplified by the lumbar spine and attenuated by the thoracic spine. The amplification seen at the lumbar spine may help maintain bone health and explain the low incidence of vertebral fractures in that spine section (Izzo et al., 2013). Although the thoracic spine attenuated most vibration produced during physical activities, certain levels of the older thoracic spine which are prone to fracture with osteoporosis

(T8 and T12), experience the same vibration transmission as a young and healthy individual. This same vibration transmission between older and younger spines may be one of the factors that cause vertebral fractures at this level of the thoracic spine. One hypothesis is that the musculoskeletal system of the thoracic spine may not be sufficient to help the biomechanical response of the spine to prevent vertebral fractures for people with osteoporosis. The stability provided by the rib cage, sternum and thoracic kyphosis may not be sufficient for the prevention of vertebral fractures (Watkins et al., 2005, Briggs et al., 2007). Similarly, the smaller thoracic vertebrae compared to lumbar vertebrae may be a great disadvantage and a factor contributing to vertebral fractures (Tortora and Grabowski, 2003).

Spinal vibration transmission is different for different types of physical activities performed and is dependent on frequency. This may explain why the treatment of spinal osteoporosis has seen more positive results through physical activity in comparison with whole body vibration (Singh et al., 2011). Whole body vibration offers only a limited set of frequencies and magnitudes in comparison with those characteristics of physical activity.

7.4. Effect of ageing, osteoporosis and walking speed

This study was performed based on the implicit assumption that ageing and osteoporosis were, in some way, related to each other and that both have some effect on vibration transmission. Clear significant effects of ageing and osteoporosis have been identified and their frequency dependency revealed. The findings in this study have suggested that there are multiple factors that determine spinal stiffness (measured as vibration transmissibility) of the different sections of the spine. These factors are delimited by diverse anatomical, functional and viscoelastic material properties as well as walking speed. However, the results suggest that ageing may have a greater effect on vibration transmission than osteoporosis alone. This hypothesis may be tested in the future by comparing young individuals with osteoporosis with older healthy individuals.

The presence of osteoporosis decreased the lumbar vibration amplification during stair negotiation while it had a minimal effect on the thoracic spine. Ageing promoted lumbar vibration

amplification while it resulted in vibration attenuation at the thoracic spine (generally during stair descent, at specific frequencies). Although the general tendency of the thoracic spine was to attenuate vibration, findings from this study have highlighted that vibration amplification can be seen during straight walking and turning. Where vibration amplification is seen, it can be hypothesized that it is due to the stiffer tissue seen due to ageing and secondary to the loss of collagen (Shuster, 2005, Castelo-Branco et al., 1994). Although older adults are not classified with osteopenia or with osteoporosis, they may have some degree of vertebral deformity due to bone creep (Pollintine et al., 2009). Therefore it can also be hypothesized that vibration transmission is amplified due to bone creep. The amplification seen may also be due to the strong muscular contraction required to maintain posture and therefore the greater loads exerted on the spine (Izzo et al., 2013). Spinal curvatures, as an indication of posture, were measured in the present study and their relationship with vibration transmissibility was found. Dynamic spinal curvatures may be also related to vibration transmissibility.

It is unclear if the predominant amplification seen at the lumbar spine indicates that ageing has greater effects on vibration transmission in comparison with osteoporosis. Similarly, it is not clear if the increased vibration attenuation seen at the thoracic spine with ageing offers protection for osteoporotic bone or removes the mechanical stimulus necessary to stimulate bone. This latter effect of reducing the likelihood of delivering mechanical stimulus to the thoracic spine regardless the type of physical activity may be possible and rather concerning. For instance, a previous study has highlighted that there is no association between physical activity (at any level) performed during late adolescence and the vertebral strength seen at the age of 21 (Junno et al., 2013). The reduced vibration observed in this study may explain why there is no association between physical activity and vertebral strength. However, this may not apply to all regions of the spine. As presented in this study, there are significant differences in vibration transmission between lumbar and thoracic spines.

Despite the plethora of studies, organizations and institutes recommending walking as a key activity to promote the maintenance of bone and the stimulation of new bone, none provided clear evidence on the possible mechanism that favoured this activity over others. This study may suggest that during walking, the vibration magnitude delivered to the lumbar spine is least affected by ageing and osteoporosis. At different levels of the thoracic spine, ageing was generally associated with vibration of lower magnitude while osteoporosis with both increase and decrease of attenuation of vibration .Similar to the lumbar spine, the greatest vibration magnitudes are observed during walking but also during stair descent. However, it is not known if the vibration magnitude measured during walking is optimal to stimulate bone safely and effectively. The same can be said for other observations in vibration transmission at the lumbar spine. For example, there are frequencies and physical activities at which the lumbar spine receives the same percentage of vibration regardless of ageing and osteoporosis. In order to determine what this response means in terms of bone metabolism, further research is needed. At this stage is not possible to recommend a specific activity.

Physical activities deliver vibration to the lumbar and thoracic spine in significantly different ways. The way this vibration is delivered to the lumbar and thoracic spines is also dependent on frequency, achieving at some instances the same response between lumbar and thoracic spines. This is important to clinicians who are in charge of prescribing physical activity to people with osteoporosis. It is recommended that prescribed physical activity, as part of a healthy lifestyle or as a treatment for osteoporosis, must take into account the differences in the mechanical response between lumbar and thoracic spine. As presented in this study, lumbar and thoracic spines respond differently to dissimilar physical activities. Depending on the severity of osteoporosis and the levels of the spine affected, certain types of physical activity will enhance stimulation or prevent it in both different and equal ways at lumbar and thoracic spines, depending of the frequency components produced during gait.

This study highlights the fact that ageing is associated with increased vibration transmissibility but a decreased magnitude of vibration transmitted during fast walking compared to normal walking. These opposing effects have conflicting consequences on the spine. Clinicians seem to recommend fast walking under the belief that it stimulates the spine with greater forces and under the assumption that these greater forces will stimulate bone growth (Groothausen et al., 1997). Previous studies have also suggested the use of an osteogenic index (OI) as well as bone loading units (BLU) based on the magnitude of ground reaction forces (Turner and Robling, 2003, Dolan et al., 2006). However, these indexes and units neither consider loading of specific sections of the skeleton (such as the spine) nor the implication of amplification and attenuation of the vibration transmitted through the body. Data from this study has contributed towards the first clear quantitative demonstration that the effects of fast walking on the transmission of vibration of the spine during gait are of a complex nature. The effects of fast walking are of a complex nature since they are dependent on frequency as well as on the type of physical activity performed (Morgado Ramírez et al., 2013a, Morgado Ramírez et al., 2013b). This is also the first study to show that fast walking is capable of promoting attenuation rather than amplification of vibration at specific frequencies. Although physical activity experts recommend 'brisk' walking in order to maintain BMD of the older spine (Daniels, 2008, Martyn-St James and Carroll, 2009, Langsetmo et al., 2012, UK Chief Medical Officers, 2012, Winter-Stone, 2005, WHO, 2012, NOF, 2010, NOF, 2012, NOS, 2012, Cheung and Giangregorio, 2012, Van Norman, 2010), it is not clear whether this attenuating response of the thoracic spine is protecting the older (potentially weak) spine from harmful vibration or simply preventing it from receiving the needed stimulation that would promote stronger vertebrae with consequent fracture reduction or prevention. The findings of this study also provide evidence of how the eighth thoracic vertebra (T8) may not be stimulated even during fast walking. Further work is required to investigate if the reduced stimulation delivered to T8 promotes bone growth. It is hypothesized that the stimulation received by T8 is not sufficient to promote a significant increase in bone strength due to the documented high incidence of vertebral fractures with and without osteoporosis at this level of the spine (Ravishankar, 2009). For instance,

a recent study suggested that further exercise studies that focus on vertebral fracture as an end point are needed since current evidence suggests that although exercise reduces overall fractures the extent of the reduction is reduced for vertebral fractures in older people. Thus the attenuation of vibration observed in this study may be a factor contributing to the seemingly small influence that physical activity has on vertebral fracture risk. It is strongly suggested to evaluate the clinical effects of fast walking in volumetric bone strength in order to improve previous studies which only studied the effects of fast walking on BMD (Brooke-Wavell et al., 2001, Martyn-St James and Carroll, 2008, Schmitt et al., 2009). Finally, it is possible that the effect of physical activity has an accumulative effect on bone (Qin et al., 1998) . For example, it has previously been observed that walking for 6 to 12 months does not preserve BMD in the lumbar spine of postmenopausal women (Martyn-St James and Carroll, 2008). It is necessary to investigate if by increasing walking speed, a significant change in BMD and bone strength is seen.

Data from this study has contributed towards the first clear quantitative demonstration of the frequency and magnitude components of vibration delivered to the lumbar and thoracic spines during physical activity during normal and fast walking. Spine vibration transmission during gait has not been measured before in people with osteoporosis. This study has also provided evidence that this vibration measurement technique is useful beyond young and healthy people and can be used when studying the role of osteoporosis in the older and frail population.

7.5. Significance of spinal curvatures

Important contributors of maximum transmissibility at maximum acceleration spectral density were determined through multiple regressions. Between 15.3% and 66.8% of vibration transmission variability was explained through vibration magnitude, spine curvatures, BMI, T-score, gender and walking speed. From this, between 1.5% and 10.7% corresponded to the variance explained by spinal curvatures. These findings support previous suggestions that spinal curvatures have a strong relationship with the exerted load on the spine (Briggs et al., 2007, Bazrgari et al., 2008, Singh et al., 2011). Although not significant changes in spinal curvatures can be seen or measured (as reported in the present study) between young and older subjects (with and without osteoporosis),

lumbar lordosis and thoracic kyphosis have important and significant roles in the percentage of vibration transmitted during physical activity. This is important as it indicates an additional way of detecting important changes in vibration transmissibility through an individual spine through time. However, the association of other factors such as vertebral creep, intervertebral disc degeneration and the presence of osteoporotic fractures with vibration transmissibility still have to be established. In future work, a threshold of change of spinal curvature that could indicate significant effects in vibration transmission and consequently in the risk of fracture during physical activity could be determined. It is clear that the spinal curvatures have a relationship with vibration transmission.

7.6. General limitations

The amplification of vibration signals at the lumbar spine may be related to active contraction of muscles during physical activity. However, the extent of the effect of spinal muscular contraction on vibration transmission is unclear. It has been suggested that compressive forces increase during standing given vertebral wedging (Briggs et al., 2006). For example, vertebral fractures at the thoracic level will induce greater flexion for which extensors need to compensate by increasing the moment to counter gravitational forces and maintain an upright position (Luo et al., 2010a). Similarly, the hypothesis that the lumbar spine naturally exerts greater loads on the vertebrae due to its instability compared with the thoracic section, which is more stable due to the thorax, still need to be verified. It may be that fractures are also associated with spinal torsional forces. The measurement of vibration transmission employs a technique based on inertial sensors, which can provide acceleration data in the mediolateral and anteroposterior directions. However, as mentioned previously, the acquisition of three dimensional data through inertial sensors containing magnetometers is affected by metal interference. Inertial sensing technology has to be improved to enable its use at any location. Thus the choice of the vertical acceleration may simplify the calculation of vibration transmission as well as aid in the employment of current inertial sensing technology for daily monitoring.

The effect of ageing and osteoporosis on vibration transmission during physical activity takes into account all types of tissue that integrate the human spine. It is not possible to separate specific tissue mechanical effect, for example those related to bone alone. Ageing and osteoporosis affect not only bone but surrounding soft tissue, causing profound changes to the mechanical response of the spine. Yet, the vibration transmission measurement technique employed enables the characterization of the overall spinal mechanical response (with all its various tissues interacting) during gait.

The vibration transmission technique is able to detect the effects due to ageing and osteoporosis but is difficult to ascertain which factor contributes the most to the vibration amplification and attenuation seen. It has been suggested that vertebral bone is deformed with time (creep) (Pollintine et al., 2009). So if damage due to creep accumulates with age, this could justify the vibration transmission changes seen due to ageing and in absence of osteoporosis. Similarly, it is not clear if the hypothesized creep due to ageing has greater effects on vibration transmission in comparison with osteoporosis. This in turn raises the possibility that osteoporosis induces a similar creep effect as normal ageing and then it is more difficult to determine isolated effects of osteoporosis in the older population. It is not possible to know if any of the older healthy volunteers of this study was having vertebral creep. It is not possible to determine creep from vibration transmission in this study. Future work may be able to measure vibration through the technique presented here and quantify creep of vertebrae and intervertebral discs through other methods (fluoroscopic imaging, MRI) (Wang et al., 2009) simultaneously to determine their relationship. In line with this thought, any comparison between the young and healthy and older osteoporotic groups presented in this thesis is a mere observation of the mechanical response of the spine during physical activity. Its clinical value cannot be determined with this thesis, future research is necessary. Therefore, any observation produced from comparing young and healthy and older osteoporotic subjects is presented with caution and cannot be disregarded the clinical implications are yet to be determined.

The concept of vibration transmission consists in transmissibility (percentage vibration transmitted), frequencies and a magnitude, which represents the complexity of the system studied (biomechanics of the human spine). There are many ways of interpreting the mechanical response of the spine during physical activity. In order to understand how ageing, osteoporosis, vibration transmission and spinal curvature are related to vertebral fractures and bone metabolism further research is required. Thus, one limitation of this study is that it is not known how bone cells may respond to the vibration produced during physical activity. It could be argued that osteoporosis disables the bone cells' response to stimulation therefore limiting the scope of physical activity as a treatment for osteoporosis. However, further research is necessary to understand the effect of osteoporosis at cellular level (Thompson et al., 2012). Similarly, this study did not measure the number of cycles delivered during each trial. It has been suggested previously that bone has a nonlinear response to loading intensity and number of cycles (Qin et al., 1998). Thus the results presented in this study must be used carefully as they represent vibration percentage and magnitude measured during only one session of exercise on the same day for each participant.

7.7. Future work

Unfortunately, for the frail population that cannot perform physical activity (due to illness or disability) stimulation to bone has to be delivered through another technique, for example, through whole body vibration (WBV). The frequencies and magnitudes produced during physical activity are different to those produced by vibrating force plates. WBV manufacturers focus on high frequency and low magnitude (Thompson et al., 2012). Future research is needed to determine if vibration produced during physical activity is more anabolic to bone than WBV, if so, it will be necessary to redesign current WBV devices to reproduce the frequency and magnitude components produced during gait.

Before this research study, it was not possible to tell with certainty the mechanical response of the lumbar and thoracic spines to physical activity, particularity with older healthy and osteoporotic people. Transmissibility and RMSa measurement have provided evidence of their capability to reliably measure the magnitude, frequency and percentage of vibration transmitted to the human

spine during daily life activities. This offers an exciting tool for research and development as well as for clinical use. Future work applications and impacts have been delineated in three stages (Table 7.7-1). Stage 2 and 3 require the partial or full completion of stage 1 in order to achieve the greatest benefits for the population suffering from osteoporosis. In stage 1, the biological effects of this transmitted vibration should be established. For instance, one important research question will be to ask if the stimuli delivered through physical activity affects bone growth. Short and long term studies could employ transmissibility to reliably document the response of healthy and osteoporotic spines to different types of physical activity and to different durations of stimulation (with different number of cycles). It has been suggested that bone response is sensitive to the frequency of the stimulus (Thompson et al., 2012, Rubin et al., 2001). Thus mechanical stimulation will influence bone response according to the number of loading cycles as well as according to how often a set of cycles is delivered. It is desirable to develop a set of inertial sensors and software to facilitate the technology to researchers and clinicians. In the first stage, low weight accelerometers and gyroscopes could be integrated with a data logger. The miniaturization of inertial sensors will play an important part to enable the interpretation of transmissibility at wider frequencies (greater than 20 Hz) if required.

Further research is needed to determine if bone mechanotransduction signals are dependent on the amplitude of vibration being delivered to bone (Table 7.7-1). Judex and Rubin (2010) suggested that if non uniform vibration produce non uniform strain in bone, this could be spatially correlated with mechanotransduction signals and thus it could be argued that bone cells sense vibration during this type of stimulation. This thesis has provided a technique that could be employed to identify possible correlation between vibration characteristics with mechanotransduction signals in order to improve our knowledge of how the healthy and older human skeleton respond to physical activity.

Table 7.7-1 Aplication and impact of vibration transmissibility measurement

Stage	Application		Impact
1	Research and Development	Test different physical activities in populations with diseases affecting the metabolism and structural integrity of the musculoskeletal system of the spine.	Standard measurement of physical activity for treatment of musculoskeletal diseases. Correlate mechanotransduction signals with physical activity. Correlate biomarkers of bone metabolism with physical activity. Correlate transmissibility with nonlinear indicators such as multivariate sample entropy and Lyapunov exponents.
2	Clinical use / clinical trials	Acknowledge individual response to mechanical stimuli through physical activity.	Evidence based treatment of osteopenia and osteoporosis through physical activity. Evidence based complementary treatment (physical activity + pharmacological treatments + diet) of osteopenia and osteoporosis. Reliable monitoring of individual effectiveness of physical activity for the treatment of osteopenia and osteoporosis. Modification of prescribed physical activity for the treatment of osteopenia and osteoporosis according to individual response in short and long term. Evidence based prescription of physical activity for the prevention of osteopenia and osteoporosis. Evidence based short and long term effects of spinal surgeries such as vertebroplasty and kyphoplasty on vibration transmissibility.

Stage	Application	Impact
3	Policy in public health and standards	Work with the world health organization and the European union to establish a standard procedure to prescribe physical activity throughout life for prevention and treatment of osteoporosis (policy level).	Standard guidance for healthcare providers on the prevention and treatment of osteoporosis through physical activity, supported by the Department of Health. Funding of research for the development and integration of healthcare monitoring technology and telemedicine which will potentially prevent vertebral fractures or detect them in real time. Establishment of sanctions to entities that act against public health through the promotion of physical inactivity. Public campaigns against habits that reduce the impact of physical activity on the prevention and treatment of osteoporosis (potentially alcohol and tobacco consumption). Achieve healthier longevity.

Due to the nonlinear nature of vibration transmissibility it is very likely that its time behaviour can be scored through indicators from nonlinear time series analysis theory, such as multivariate multiscale entropy and Lyapunov exponents (Ahmed and Mandic, 2012, Kantz and Schreiber, 2004). More than one nonlinear measure is necessary to capture the properties of the most complex signals (Kantz and Schreiber, 2004). Entropy is an indicator of complexity and Lyapunov exponents indicates how stable a system is. Biological systems tend to be more complex and stable when healthy while losing their complexity and stability with disease (Dingwell and Cusumano, 2000). Currently, no study has presented a nonlinear analysis of the effect of osteoporosis on spinal biomechanics. This study offers an unprecedented set of acceleration data which could be employed to determine the nonlinear effect of osteoporosis. If such effect is present and significant, it could be correlated with vibration transmissibility. This will increase our knowledge of the

disease of osteoporosis and aid future clinical interventions. As an example, acceleration data (filtered at 20 Hz and corrected for skin movement) of all subjects from this study was used to calculate sample entropy through Matlab and a validated nonlinear time series analysis package called TISEAN (Hegger et al., 2007). Then acceleration data derived from the sensor over T1 and from three selected participants, each one representing a group (YH, OH and OO) was employed to calculate Lyapunov exponents through Matlab and TISEAN. Sample entropy results suggested that ageing and osteoporosis introduce complexity to human gait rather than decreasing it (Table 7.7-2). This contradicts the hypothesis that disease decreases the complexity of biological systems. Thus, it is unclear how the natural process of ageing as well as its combination with osteoporosis may result in more complex control mechanisms in human walking.

Table 7.7-2 Sample entropy for all subjects and Lyapunov exponents for three subjects, walking in a straight line at a normal walking speed

Group	Sample entropy‡	Lyapunov Exponent†
Young and healthy (YH)	2.15 (0.18) *OH *OO	0.025
Older healthy (OH)	2.33 (0.21)	0.024
Older osteoporotic (OO)	2.28 (0.25)	0.030

† Calculated for only 1 subject per group and for the sensor over the first thoracic vertebra only, for one trial,

‡ Calculated for all subjects, all trials and all locations over the spine. Mean (SD), * significant difference

The mechanisms that control human walking respond to internal and external changes in biological parameters and interactions among them (Goldberger et al., 2000). Internal biological parameters that can be altered by ageing and osteoporosis are, for example, a change in soft tissue and bone stiffness. External change parameters may involve spinal curvature and increased loads over the spine due to muscular contraction trying to resist excessive thoracic flexion. According to the theory of Lyapunov exponents applied to biological signals, the YH Lyapunov exponent was

expected to be higher than the OH and even higher than that for the OO. However, it appears that ageing and osteoporosis may not reduce the nonlinear nature of human walking, but rather increase it (Table 7.7-2). This in turn contradicts the hypothesis that ageing and osteoporosis reduce the stability of human walking, thus inducing falls. So if ageing and osteoporosis do not affect ability of human walking to respond to perturbations (dynamic stability), what is causing falls? Further research is clearly needed in this field.

7.7.1. Clinical implications

It has been suggested that finding an optimal exercise intervention for the senescent skeleton may not be sufficient for the treatment of osteoporosis (Srinivasan et al., 2011). Therefore any study, evaluating the effect of physical activity and exercise on bone, should examine the impact of leisure and non leisure related activities (Table 7.7-1). It may be that physical activities beyond the ones measured contribute to bone mineral content and structure (Shedd et al., 2007). A standard 24 hours 7 days a week of characterizing physical activity based on inertial sensors may be useful for this purpose.

Given a portable and integrated device to measure transmissibility in vivo at all times, it will be possible to determine dynamic boundaries in order to alert patients and doctors of possible vertebral fractures. Evidently, these dynamic boundaries must be determined before an alert system can be developed and tested. Knowledge of tolerable limits for human exposure to vibration produced during walking is necessary to maintain health and performance in the population afflicted with osteoporosis. This will require the use of biodynamic models as well as anthropometric manikins of the spine (Brammer and Peterson, 2003). Given a biodynamic model of the spine during physical activity, it will be possible to calculate simpler measures that could be applied during daily monitoring. For example, a measurement such as vibration exposure or dose value (Mansfield, 2005a, Brammer and Peterson, 2003) could be combined with fracture risk indicators calculated through the WHO fracture risk assessment tool (FRAX®) (Kanis et al., 2009). This will provide an unprecedented data base which will increase our understanding of sudden vertebral fractures during physical activity. A limitation of this direction is the development of the

boundaries as, it is not possible to expose people with osteoporosis to exercise routines that will potentially cause vertebral fractures due to ethical reasons.

An additional future application of vibration transmissibility measurement involves experts in surgical intervention of the spine (Table 7.7-1). After a vertebroplasty or a kyphoplasty, clinicians could benefit by understanding short and long term effects of surgery on spinal vibration transmissibility. This means that surgeons will be able to reliably determine the effectiveness of the surgery in terms of returning the spine to a healthy biomechanical response apart from repairing current vertebral fractures. The ideal spinal surgery would return the spinal biomechanical response as close as possible to a healthy, age-matched spine. If the response is far from a healthy one, further measures would need to be taken. Vibration transmissibility measurement may also enable clinicians to monitor the progress of the biomechanical response of the patient's spine after surgery. This could give information on how the spine adapts to surgery over the short and long term, in biomechanical terms. Finally, clinicians may also be able to provide recommendations to patients that had undergone spinal surgery on the type of physical activity which could be performed safely while maintaining or increasing bone strength. Therefore future research should also focus in the employment of vibration transmissibility in the study and monitoring of current surgical interventions of the healthy and osteoporotic spine.

Once the relationship between transmissibility and strength changes in bone are determined, the technique presented in this study offers the opportunity to establish standard procedures to prescribe physical activity (Table 7.7-1). This will provide the information to health institutions and organizations to promote public health policies to manage osteoporosis. Currently physical activity does not play a clearly defined role in managing osteoporosis due to the lack of evidence regarding the impact of physical inactivity.

7.8. Final conclusion

Surface measurement of vertical vibration transmissibility of the spine during physical activities is possible with the correction method presented. Vibration transmitted to the lumbar and thoracic spines was significantly affected by the type of physical activity performed and dependant on age, BMD, spinal curvature, BMI, vibration magnitude and walking speed. The lumbar spine generally amplifies vibration while the thoracic spine generally attenuates it. During fast walking the lumbar spine preserved its amplification tendency and the thoracic spine its attenuation tendency.

Ageing and osteoporosis affect the vibration transmission of the spine in different ways. Ageing is generally associated with vibration amplification. Osteoporosis decreases vibration amplification of the lumbar spine during stair negotiation. Walking was the physical activity least affected by ageing and osteoporosis. It is unclear if the stimulus received by the human spine during walking can stimulate bone growth.

Even though the thoracic spine attenuates vibration, locations prone to fracture transmit the same vibration as healthy and young individuals during specific physical activities. This highlights the need to characterize the effect of prescribed physical activity in vivo for safety purposes. By increasing the walking speed the amplification and attenuation tendency of the lumbar and thoracic spines respectively, is accentuated rather than altered. Fast walking may not necessarily increase the stimulus delivered to the thoracic spine.

Differences in vibration transmission between lumbar and thoracic sections during normal and fast walking speeds emphasize the need to rethink current physical activity prescription. Physical activity experts must consider these differences in order to stimulate different sections of the spine safely and effectively.

Future research should examine the optimal dose of mechanical stimulus (as determined by the magnitude, frequency and percentage transmission of such vibration) required for stimulating bone growth.

Finally, this thesis has significantly added to our understanding of the characteristics of vibration produced during physical activity and transmitted through the human spine. This is pivotal in understanding the effect of ageing and osteoporosis on the way physical activity may or may not stimulate bone growth at the spine. The mechanical characteristics of different regions of the spine may be related to the incidence of vertebral fractures. This thesis has taken an important first step to provide information to support future studies on the determination of the optimal physical activity dose for people with and without osteoporosis. Further research should employ this vibration measurement technique in randomized controlled trials to identify the intensity and types of physical activities that significantly increase the risk of vertebral fractures in people with osteopenia and osteoporosis while taking into account the different mechanical response of the lumbar and thoracic sections. It is hoped that the knowledge acquired will lead to a significant breakthrough in the way physical activity is prescribed which will help millions of older adults with osteopenia and osteoporosis.

Appendix A Ethics documentation

A1 Ethical approval letter

Roehampton University London

Research & Business
Development Office

Roehampton University
Froebel College
Roehampton Lane
London SW15 5PJ

Tel: 020 8392 3300
Tel: 020 8392 3554 (direct line)
Fax: 020 8392 3550
Email: business.services@roehampton.ac.uk
www.roehampton.ac.uk

10th September 2010

Ms Dafne Zuleima Morgado Ramirez
111 Sherfield Gardens
Putney
London
SW15 4PA

Dear Dafne,

Ethics Application (research student)
Applicant: Dafne Zuleima Morgado Ramirez
Title: Does physical activity affect the risk of vertebral fractures in older adults with osteoporosis?
School: HALS

Please accept this letter as confirmation that the above ethics application was approved by the Ethics Committee on 10th August 2010. We do not require anything further in relation to your Ethics Application and you may commence your research.

Many thanks,

Lemady Rochard
Research Policy Officer
Research and Business Development Office
208 Grove House, Froebel College
Roehampton University
Roehampton Lane
London
SW15 5PJ

T: +44 (0)20 8392 3256
E: L.Rochard@roehampton.ac.uk

A2 Information sheet

Roehampton University London

INFORMATION SHEET FOR PARTICIPANTS

Research Title:
Does physical activity affect the risk of vertebral fractures in older adults with osteoporosis?

Invitation Paragraph
We would like to invite you to participate in this research project. Before you decide it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information. Take time to decide if you wish to take part.

It is up to you to decide whether or not to take part. If you do decide to take part, you will be given this information sheet to keep and asked to sign the attached consent form to say that you agree and understand what this study is about. You are free to withdraw at any time and without giving reason, you can do this using a participant number that will be assigned and given to you through a Debriefing Form.

Thank you for reading this.

What is the purpose of the study?
The aim of this study is to investigate the physical properties of the spine and stability of the body in people with and without osteoporosis. These measurements would be made by sensors attached to your back while you perform simple movements that you would normally do in your everyday life.

Who have we asked to participate?
One hundred female and male healthy adults (aged 25 to 35), healthy older adults (aged 55 and above) and older adults with osteoporosis (aged 55 and above) from the general population.

Who must we exclude?
Unfortunately, we ask you not to participate if you had back or leg pain in the last 12 months that required medical treatment. Also if you have tumours, severe rheumatological disorders, tuberculosis or an infection in your spine and if you had a dislocation or surgery of the spine and lower limbs at any time of your life. You will not be able to participate if you have any history of previous osteoporotic fractures on your spine or lower limbs and if you are pregnant or allergic to ultrasound gel or adhesive tape.
We will ask you not to participate if you have an orthopaedic implant (a medical device that replaces part or a whole joint) or an electrically powered medical implant (for example a pacemaker, an implantable defibrillator, a cochlear implant, neurostimulators or an insertable cardiac monitor).
Finally, we are unable to let you participate if you have been clinically diagnosed with obesity.

When and where will the study take place?
The study will take place at a time that is convenient to you. It will take place inside Whitelands College of Roehampton University. The room and areas where you will be asked to participate, will be closed to common users (students and staff) while this study takes place.

How long will the study last?
It will last approximately two hours and a half.

What will happen to me if I take part?
If you decide to participate in the study you will be asked to attend on one occasion only and bring with you loose and comfortable clothing (for example loose trousers, a loose T-shirt or a tank top). This is will help us to have access to your back. There is a changing room for you to use when you arrive. You will also be asked to bring the shoes that you use most of the time.

We will ask you if you are right-handed or left-handed, your age, weigh you and measure your height.
A test to determine your bone mineral density (BMD) will be made three times for the left or right ankle. This test is called broad ultrasound attenuation test. Your ankle will be put between two transducers pads with gel. The calf of your leg will be positioned and secured with Velcro® straps to a support. All surfaces that your feet will be in contact with will be disinfected before being used. Please see figure 1 for reference.

You should not feel any pain, discomfort or increment in body temperature during the test. Any unusual sensation should be reported to the researcher.

After this a sensor will be used to record the locations shown in figure 2 and the curvature of your spine while you are standing.

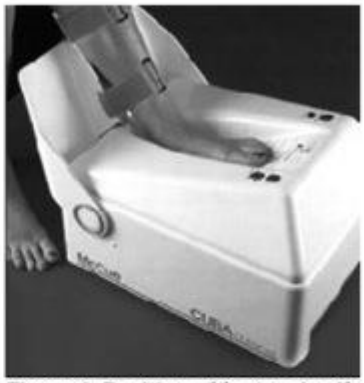

Figure 1 Position of foot and calf.

Then, small motion sensors will be secured using double sided adhesive in the three locations shown in figure 2. Skin marking and ultrasound may be used to help locate the bones of the body for sensor attachment. You should not feel any pain, discomfort or increment in body temperature during the test. Any unusual sensation should be reported to the researcher.

In order to measure the skin movement between the sensor and the underlying bone, a test will be performed four times on each sensor by displacing it approximately one centimetre by the fingers and quickly releasing it afterwards.

The physical activity that you will be requested to do is:

1. Walk along a straight line, three times.
2. Walk up and down stairs consisting of steps of normal height, three times.
3. Walk through a path consisting of combined turning and walking, three times.

You will be asked to perform the above activities at self-selected and maximum possible walking speeds.

Loads transmitted through the body will be measured using these sensors during the above activities.

Exclusively female investigators and technicians are available to perform the measurements in case you prefer this. Temperature of the room and other areas used for this study (stairs and corridor) will be kept at a comfortable temperature as possible.

Finally, the investigator in charge will detach the sensors from your back and clean any marks.

For some participants, you will be asked to do an additional test in which sensors will be attached (with double sided adhesive) to the first, third and an extra location over your spine. Skin movement will be measured with these sensors by displacing it approximately one centimetre by the fingers and quickly releasing it afterwards. This will be done four times for each sensor. Afterwards you will be asked to walk up and down stairs and along a straight line three times.

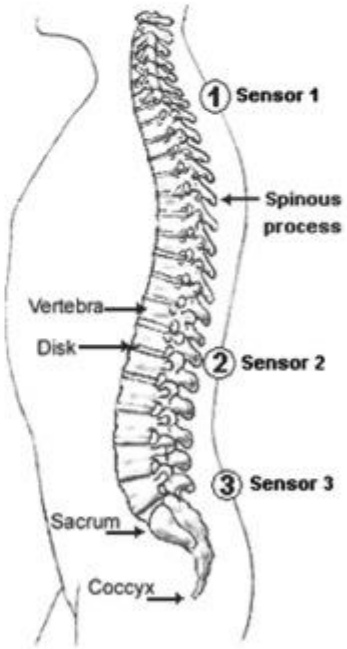

Figure 2 Parts of the spine and three points marked over it.

Are there any risks involved in participating?

This study has been designed to investigate movements that you would do in your everyday life. It involves walking in a straight line, turning and walking up and down stairs of normal height and width. The sensors and the ultrasound do not have any health risks known to the researchers. The gel and adhesive tape used should not cause any allergic reaction. If any concern or discomfort is felt either before or during the tests, please tell us.

Are there any benefits involved in participating?
You will get a free bone scan assessment that usually has a value of £20 in private clinics. At the conclusion of the project, if you agree to give us your personal contact information we will send you a description of the major findings as well as reference to research publications generated from this project.

Will we compensate you for your time?
To thank you for taking time to participate and for your inconvenience we will offer you a £10 voucher at the end of your participation. You will not have any financial benefit from the research in any other respect.

How will we maintain your privacy and confidentiality?
To preserve anonymity you will be given an identification number known only to the principal researcher and the project supervisors. Using the same number, data will be stored on a password secure computer within Roehampton University ensuring your individual information remains confidential. Once the results are reported it will not be possible to identify individual persons. After the study data will be stored in the same way for a period of ten years after which they will be destroyed.

What will happen if I don't want to carry on with the study?
You have the right to withdraw from this study at any time, without fear of prejudice. If you decide to withdraw, please tell the principal investigator at any time. If you withdraw, it may be beneficial to use your data already collected up to the point of withdrawal, but all other data will be destroyed.

Who is organising and funding the research?
This research is organised by the investigators outlined below along with the School of Human and Life Sciences, Roehampton University.
This study has been subject to full ethical review, approved and funded by Roehampton University.

	Name	University Address	Email	Telephone
***Principal* investigator**	**Dafne Zuleima Morgado Ramirez** Postgraduate student	School of Human and Life Sciences, Roehampton University, Whitelands College, Holybourne Avenue, London, SW15 4JD	d.morgado-ramirez@roehampton.ac.uk	+44 (0)20 8392 3472
Supervisor	**Raymond Lee** Professor of Biomechanics and Head of Sports Sciences		r.lee@roehampton.ac.uk	+44 (0)20 8392 3539
Co-Supervisor	**Siobhan Strike** Senior lecturer		s.strike@roehampton.ac.uk	+44 (0)20 8392 3546

What if I have questions about the project?
For information and doubts about this research study, please talk with the principal investigator.
Please note this study is being completed as part of a Doctor of Philosophy (PhD) educational qualification.

What if there is a problem or complaint?
Technical staff fully trained as first aider will be available throughout your participation.
If there is a problem at any time and you would like to contact someone then your contact should be:

	Name	University Address	Email	Telephone
Director of Studies	**Raymond Lee**	School of Human and Life Sciences, Whitelands College, Roehampton University, Holybourne Avenue, SW15 4JD	r.lee@roehampton.ac.uk	+44 (0)20 8392 3539

If you are a student in Roehampton University and feel any physical or emotional discomfort about any aspect of the study, please contact your **Student Welfare Officer** who will be able to advise you on support groups that can deal with your particular concern:

College	*Officer*	*Telephone*
Frobel	Anne-Marie Joyes	+44 (0)20 8392 3304
Digby Stuart	Jo Granger	+44 (0)20 8392 3204
Southlands	Belinda Scott	+44 (0)20 8392 3402
Whitelands	Ejiro Ejoh	+44 (0)20 8392 3502

If you feel that your concerns are more serious or complex please contact the **Student Medical Centre** on +44 (0)20 8392 3679. If you are not a student at Roehampton University, please contact your nearest **General Practitioner**.

If you need this information sheet in large printing, please request it from the principal investigator.

A3 Check list

CHECK LIST to be completed by the investigator

Does physical activity affect the risk of vertebral fractures in older adults with osteoporosis?

Date: _____ /_____ /_____

Investigator:

Subject ID number: ☐☐☐

EXCLUSION CRITERIA

- ☐ Back or leg pain in the last 12 months that required medical treatment
- ☐ Dislocation of the spine or lower limbs
- ☐ Surgery of spine or lower limbs
- ☐ Rheumatological disorders
- ☐ Previous osteoporotic fractures
- ☐ Orthopaedic and/or electrically powered medical implant
- ☐ Tumors
- ☐ Pregnancy
- ☐ Tuberculosis
- ☐ Spinal infection
- ☐ Allergic to ultrasound gel and adhesive tape
- ☐ Prefers a male investigator to perform measurements
- ☐ Other muscular-skeletal injuries, disease or limitations that may affect natural gait

GENDER

- ☐ Female
- ☐ Male

AGE

- ☐ 25-35
- ☐ 55 and above

DOB ____/____/____

WEIGHT

HEIGHT

HANDEDNESS

- ☐ Right handed
- ☐ Left handed

OSTEOPOROSIS

- ☐ Not osteoporotic
- ☐ Osteoporosis
- ☐ Osteopenia
- ☐ Don't know

HABITS

- ☐ Smoking
- ☐ Drinking

Contact information for debriefing:

NOTES

BUA TEST

Left	Right

A4 Consent form

Roehampton University London

PARTICIPANT CONSENT FORM

Please complete this form after you have read the Information Sheet and/or listened to an explanation about the research.

Title of Research Project:

Does physical activity affect the risk of vertebral fractures in older adults with osteoporosis?

Brief description of Research Project:

Study the mechanical properties of the spine and stability of the body in people with and without osteoporosis. Measurements would be done while you perform simple movements that you would normally do in your everyday life. Please refer to the Information Sheet for details.

Consent Statement:

I understand that I can notify the investigators involved if I decide at any time during the research that I no longer wish to participate in this project, and withdraw from it immediately without fear of prejudice.

I consent to the processing of my personal data for the purposes of this study. I understand that such information will be treated as confidential and that my identity will be protected in the publication of any findings. If I decide to withdraw, I understand that it may be beneficial to use my data already collected up to the point of withdrawal but all other data will be destroyed.

I agree that the research project named above has been explained to me to my satisfaction and I agree to participate in it. I have read this consent form and the Information Sheet and understand what the research study involves.

Name .....

Signature ..... Date .....

I **Dafne Morgado** confirm that I have carefully explained the nature, discomfort and any foreseeable risks (where applicable) of the proposed research to the volunteer.

Signature ..... Date .....

1

If you have a concern about any aspect of your participation or any other queries, please raise this with the investigator.

	Name	University Address	Email	Telephone
***Principal* investigator**	**Dafne Zuleima Morgado Ramirez** Postgraduate student	School of Human and Life Sciences, Roehampton University, Whitelands College, Holybourne Avenue, London, SW15 4JD	d.morgado-ramirez@roehampton.ac.uk	+44 (0)20 8392 3472
Supervisor *Head of Life Sciences Department and Director of Studies*	**Raymond Lee** Professor of Biomechanics and Head of Sports Sciences		r.lee@roehampton.ac.uk	+44 (0)20 8392 3539

A5 Debriefing Form

Roehampton University London

DEBRIEFING FORM

Participant ID Number: __ __ __

Title of Research Project:

Does physical activity affect the risk of vertebral fractures in older adults with osteoporosis?

Brief Description of Research Project:

The aim of this study was to investigate the mechanical properties of the spine and stability of the body in people with and without osteoporosis. These measurements were made by sensors attached to your back while you performed simple movements that you would normally do in your everyday life.

Principal Investigator Contact Details:

Name: Dafne Zuleima Morgado Ramirez, PhD student

Address: School of Human and Life Sciences, Roehampton University, Whitelands College, Holybourne Avenue, London, SW15 4JD

Email: d.morgado-ramirez@roehampton.ac.uk

Telephone: +44 (0)20 8392 3472

Your right to withdraw

You have the right to withdraw from this study at any time, without fear of prejudice. Please tell the principal investigator at any time using your ID number. If you withdraw, it may be beneficial to use your data already collected up to the point of withdrawal, but all other data will be destroyed. The data that have been collected will only be securely retrieved and analysed using the participant ID number assigned to your data set. Your anonymity will be maintained at all times.

Further information and contact details

For further information about this research project, please feel free to contact Dafne Morgado (Tel: +44 (0)20 8392 3472, email: d.morgado-ramirez@roehampton.ac.uk) or the Director of Studies: Raymond Lee (Tel: +44 (0)20 8392 3539, email: r.lee@roehampton.ac.uk)

Please turn over

1

If you have a concern about any aspect of your participation or any other queries, please raise this with the investigator.

If you are a student in Roehampton University and feel any physical or emotional discomfort about any aspect of the study, please contact your **Student Welfare Officer** who will be able to advise you on support groups that can deal with your particular concern:

College	*Officer*	*Telephone*
Frobel	Anne-Marie Joyes	+44 (0)20 8392 3304
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Southlands	Belinda Scott	+44 (0)20 8392 3402
Whitelands	Ejiro Ejoh	+44 (0)20 8392 3502

If you feel that your concerns are more serious or complex please contact the **Student Medical Centre** on +44 (0)20 8392 3679. If you are not a student at Roehampton University, please contact your nearest **General Practitioner**.

Thank you for participating in this research, we greatly appreciate your contribution.

Appendix B Subjects

Table Appendix 1 Final sample size for each test and group

		Walking speed and physical activity							
YH (34)		NWS				FWS			
		w	m	a	d	w	m	a	d
RMSa	T1	34	34	34	34	34	34	34	34
	T8	29	29	29	29	29	29	29	29
	T12	33	33	33	33	33	33	33	33
	S1	33	33	33	33	34	33	33	33
T	Lumbar	31	32	32	32	32	32	32	32
	Thoracic	33	32	34	34	34	32	34	34
OH (23)		NWS				FWS			
		w	m	a	d	w	m	a	d
RMSa	T1	23	23	23	23	23	23	23	23
	T8	21	21	21	21	21	21	21	21
	T12	21	21	21	21	21	21	21	21
	S1	23	23	23	23	23	23	23	23
T	Lumbar	19	19	19	19	19	19	19	19
	Thoracic	23	23	23	23	23	23	23	23
OO (43)		NWS				FWS			
		w	m	a	d	w	m	a	d
RMSa	T1	43	43	43	43	43	43	43	43
	T8	41	41	41	41	41	41	41	41
	T12	38	38	38	38	38	38	38	38
	S1	43	43	43	43	43	43	43	43
T	Lumbar	38	38	38	38	38	38	38	38
	Thoracic	42	42	42	42	42	42	42	42

Root mean square acceleration (RMSa), young and healthy (YH), older healthy (OH), older osteoporotic (OO), first thoracic vertebra (T1), eighth thoracic vertebra (T8), twelfth thoracic vertebra (T12), first sacral vertebra (S1), mean maximum transmissibility (T), straight walking (w), walking and turning (m), stairs ascent (a), stairs descent (d), self selected normal walking speed (NWS) and self selected fast walking speed (FWS)

Reasons for exclusion of data are: noisy data due to sensor touching subjects clothing, excessive skin dryness and excessive fat content of skin so skin correction was not possible to calculate, data collected was incomplete due to communication problem between the wireless inertial sensor and the receiver or subject did not perform the trial.

Appendix C Transmissibility and frequency

C1 Lumbar spine (frequency and walking speed)

Table Appendix 2 Effect of walking speed on mean frequencies (Hz) at which mean maximum transmissibility was found for the lumbar spine

Group	Activity	Mean frequency at which mean maximum transmissibility was found at frequency intervals (Hz)							
		0.5≤ f ≤2		2< f ≤4		4< f ≤6		6< f ≤8	
		NWSFWS		NWSFWS		NWSFWS		NWSFWS	
YH	w	1.38(0.43) ≈ 1.38(0.41)	=	2.54(0.43) < 2.70(0.47)*	+	4.90(0.52) < 5.13(0.60)*	+	7.40(0.59) > 7.05(0.67)*	-
	m	1.58(0.40) ≈ 1.67(0.40)	=	2.57(0.66) ≈ 2.56(0.46)	=	4.56(0.47) < 5.03(0.59)*	+	7.33(0.61) > 7.53(0.57)*	-
	a	1.52(0.34) > 1.34(0.42)*	-	2.37(0.44) < 3.20(0.79)*	+	4.97(0.63) ≈ 4.95(0.60)	=	7.50(0.53) > 7.01(0.83)*	-
	d	1.40(0.39) ≈ 1.32(0.42)	=	2.88(0.71) < 3.34(0.75)*	+	4.58(0.51) < 5.22(0.55)*	+	7.31(0.65) ≈ 7.18(0.74)	=
OH	w	1.38(0.45) ≈ 1.45(0.39)	=	2.56(0.50) < 2.80(0.55)*	+	5.18(0.47) ≈ 5.16(0.59)	=	7.47(0.61) > 7.01(0.68)*	-
	m	1.70(0.34) < 1.80(0.30)*	+	2.68(0.65) ≈ 2.54(0.41)	=	5.02(0.59) ≈ 5.09(0.59)	=	7.30(0.55) ≈ 7.23(0.68)	=
	a	1.44(0.36) ≈ 1.31(0.42)	=	2.40(0.54) > 3.36(0.66)*	-	5.34(0.49) > 4.94(0.58)*	-	7.38(0.59) > 6.99(0.77)*	-
	d	1.43(0.39) ≈ 1.43(0.39)	=	2.79(0.74) ≈ 3.35(0.75)	=	5.20(0.53) ≈ 5.19(0.52)	=	6.95(0.74) ≈ 7.20(0.69)	=
OO	w	1.44(0.43) ≈ 1.46(0.42)	=	2.63(0.57) < 2.85(0.54)*	+	5.07(0.55) ≈ 5.14(0.59)	=	7.31(0.62) > 6.91(0.70)*	=
	m	1.60(0.42) < 1.79(0.32)*	+	2.64(0.60) ≈ 2.53(0.45)	=	5.02(0.61) ≈ 5.06(0.59)	=	6.97(0.69) < 7.21(0.66)*	=
	a	1.46(0.37) > 1.31(0.41)*	-	2.66(0.71) < 3.33(0.70)*	+	5.18(0.54) > 4.92(0.58)*	-	6.97(0.73) > 6.69(0.75)*	-
	d	1.35(0.39) ≈ 1.42(0.40)	=	2.68(0.71) < 3.02(0.79)*	+	4.95(0.60) ≈ 5.06(0.57)	=	6.84(0.73) ≈ 6.93(0.78)	+

(-) decrease, (+) increase or (=) no significant change in frequency from normal (NWS) to fast walking speed (FWS), mean (SD), *=significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair descent (d), stair ascent (a)

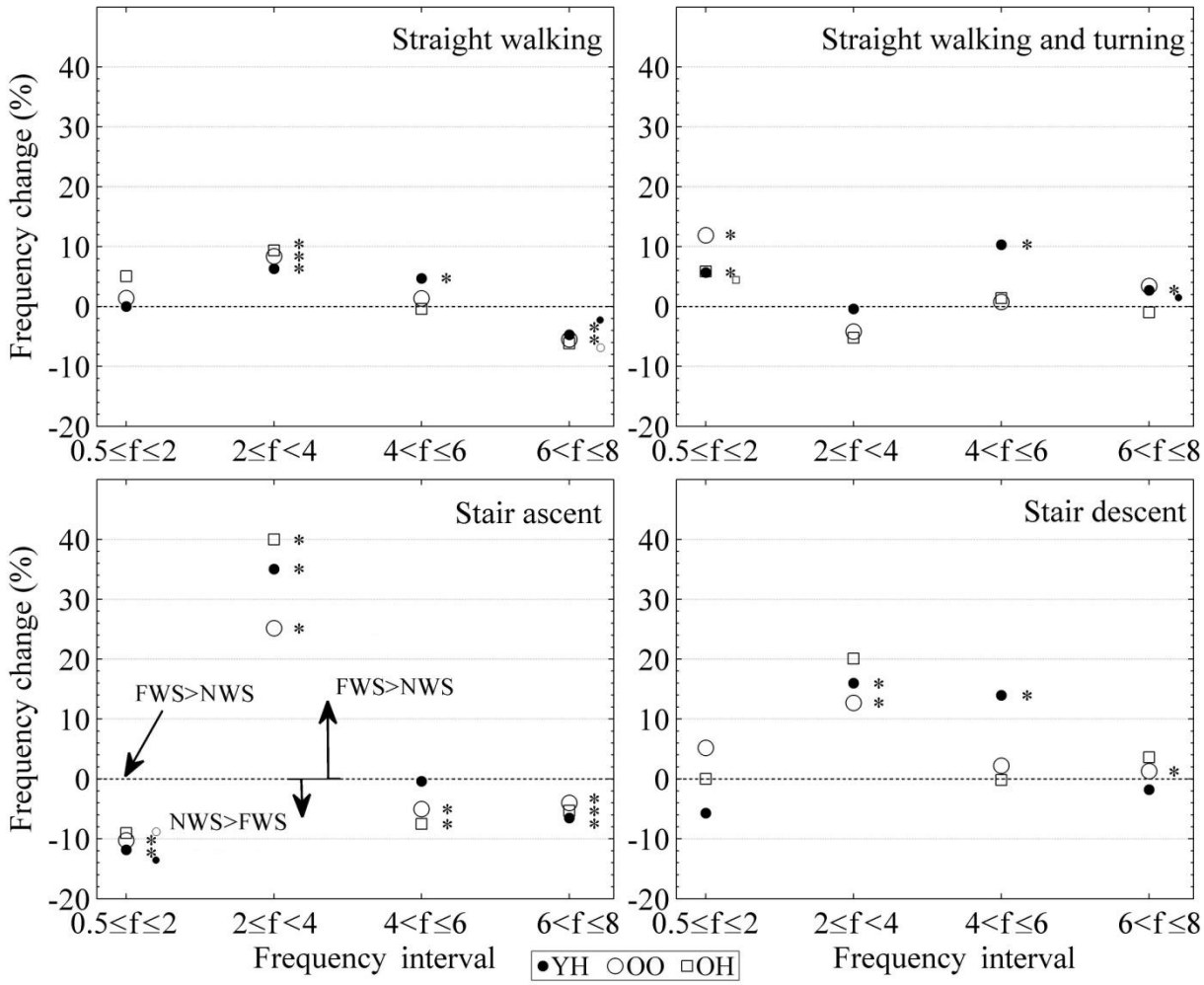

Figure Appendix 1 Relative percentage change in mean frequency at which maximum transmissibility was found for the lumbar spine from normal (NWS) to fast walking speed (FWS). *= significant change, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

C2 Thoracic spine (frequency and walking speed)

Table Appendix 3 Effect of walking speed on mean frequencies (Hz) at which mean maximum transmissibility was found for the thoracic spine

Group	Activity	Mean frequency at which mean maximum transmissibility was found at frequency intervals (Hz)							
		0.5≤ f ≤2		2< f ≤4		4< f ≤6		6< f ≤8	
		NWSFWS		NWSFWS		NWSFWS		NWSFWS	
YH	**w**	0.91(0.32) < 1.04(0.34)*	+	3.38(0.55) ≈ 3.44(0.61)	=	4.71(0.64) < 5.01(0.60)*	+	6.68(0.67) ≈ 6.74(0.64)	=
	m	0.73(0.18) < 0.86(0.29)*	+	3.19(0.57) ≈ 3.30(0.51)	=	4.81(0.61) < 5.01(0.67)*	+	6.61(0.60) < 6.83(0.63)*	+
	a	1.05(0.37) < 1.15(0.40)*	+	3.50(0.64) > 2.92(0.79)*	-	5.02(0.54) ≈ 4.83(0.59)	=	6.65(0.74) ≈ 6.65(0.77)	=
	d	1.12(0.38) < 1.27(0.42)*	+	2.69(0.74) ≈ 2.60(0.68)	=	4.78(0.53) > 4.62(0.55)*	-	6.79(0.81) > 6.54(0.77)*	-
OH	**w**	0.84(0.15) < 0.96(0.23)*	+	3.30(0.51) < 3.52(0.51)*	+	5.16(0.64) ≈ 5.07(0.56)	=	6.50(0.64) < 6.76(0.64)*	+
	m	0.71(0.11) < 0.78(0.15)*	+	3.24(0.52) < 3.46(0.43)*	+	5.14(0.62) ≈ 5.00(0.67)	=	6.33(0.47) < 6.69(0.64)*	+
	a	1.01(0.33) < 1.16(0.40)*	+	3.49(0.65) > 2.81(0.76)*	-	5.02(0.58) ≈ 4.95(0.59)	=	6.54(0.72) < 6.70(0.70)*	+
	d	1.11(0.35) ≈ 1.15(0.39)	=	2.85(0.78) ≈ 2.69(0.72)	=	4.95(0.58) ≈ 4.85(0.58)	=	6.64(0.66) ≈ 6.49(0.69)	=
OO	**w**	0.83(0.20) < 0.95(0.23)*	+	3.35(0.50) < 3.48(0.54)*	+	5.04(0.64) ≈ 5.08(0.59)	=	6.61(0.63) ≈ 6.77(0.66)	=
	m	0.69(0.11) < 0.78(0.18)*	+	3.04(0.48) < 3.36(0.44)*	+	5.08(0.53) ≈ 5.13(0.66)	=	6.46(0.55) < 6.62(0.59)*	+
	a	1.01(0.32) ≈ 1.09(0.37)	=	3.44(0.69) > 2.57(0.67)*	-	5.05(0.52) > 4.97(0.62)	-	6.62(0.74) ≈ 6.80(0.77)	=
	d	1.01(0.33) ≈ 1.04(0.34)	=	2.81(0.80) ≈ 2.68(0.76)	=	4.94(0.56) < 4.78(0.58)*	+	6.58(0.71) ≈ 6.54(0.69)	=

(-) decrease, (+) increase or (=) no significant change in frequency from normal (NWS) to fast walking speed (FWS), mean (SD), *=significant difference, young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair descent (d), stair ascent (a)

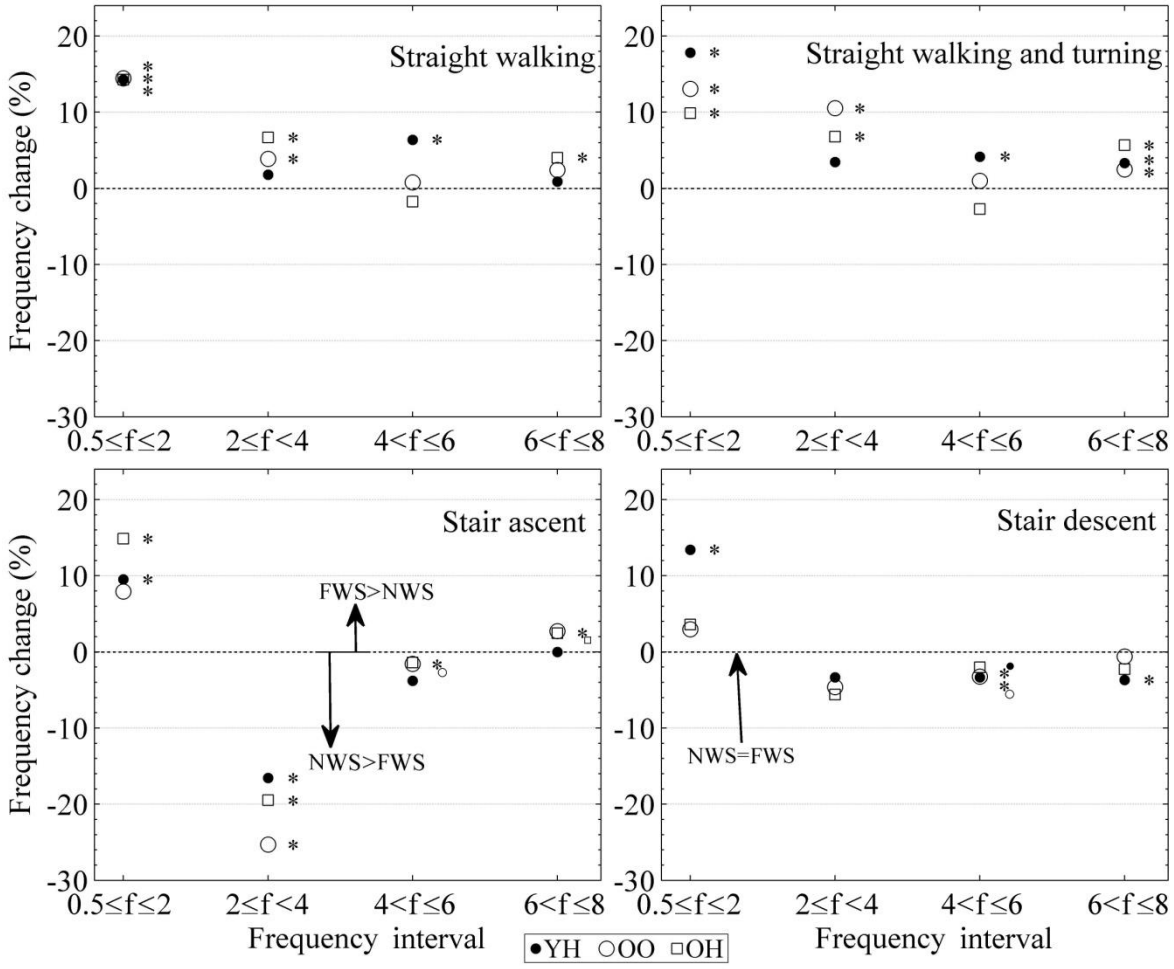

Figure Appendix 2 Relative percentage change in mean frequency at which maximum transmissibility was found for the thoracic spine from normal (NWS) to fast walking speed (FWS). *= significant change, young and healthy (YH), older healthy (OH), older osteoporotic (OO)

C3 Lumbar versus thoracic spine NWS (transmissibility and frequency)

Table Appendix 4 Significant differences betweenlumbar and thoracic spinefor maximum transmissibility at maximum acceleartion PSD and mean frequency at which transmissibility was found.All groups during normal walking speed

Group	Physical Activity	maxT@maxPSD (%)			Frequency at which maxT@maxPSD was found (Hz)		
		l	t	% difference	l	t	% difference
YH	w	114(21) > 70(15)*		-39	1.9(0.1) ≈ 1.9(0.1)		0
	m	124(23) > 61(10)*		-51	1.8(0.1) ≈ 1.8(0.1)		0
	a	124(26) > 54(10)*		-56	2.0(0.2) ≈ 1.9(0.2)		-5
	d	103(13) > 67(12)*		-35	2.3(0.3) ≈ 2.3(0.3)		0
OH	w	115(14) > 59(14)*		-49	1.9(0.1) ≈ 2.0(0.2)		5
	m	126(14) > 57(13)*		55	1.9(0.4) ≈ 2.3(0.9)		21
	a	130(29) > 53(14)*		-59	2.0(0.2) ≈ 2.3(0.6)		15
	d	117(14) > 56(13)*		-52	2.2(0.3) ≈ 2.2(0.5)		0
OO	w	119(31) > 59(15)*		-50	2.0(0.1) ≈ 2.0(0.1)		0
	m	129(38) > 56(15)*		-57	1.8(0.1) ≈ 1.8(0.1)		0
	a	124(28) > 55(13)*		-56	2.0(0.4) ≈ 2.4(0.6)		20
	d	119(17) > 53(16)*		-55	2.2(0.5) ≈ 2.1(0.4)		-5

Maximum transmissibility at maximum acceleration spectral density (maxT@maxPSD), young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair descent (d), stair ascent (a), lumbar spine (l), thoracic spine (t), * = significant difference

C4 Lumbar versus thoracic spine FWS (transmissibility and frequency)

Table Appendix 5 Significant differences betweenlumbar and thoracic spinefor maximum transmissibility at maximum acceleration PSD and mean frequency at which transmissibility was found.All groups during fast walking speed

Group	Physical Activity	maxT@maxPSD (%)			Frequency at which maxT@maxPSD was found (Hz)		
		l	t	% difference	l	t	% difference
YH	w	115(22) > 66(10)*		-43	2.4(0.2) ≈ 2.4(0.2)		0
	m	133(30) > 58(11)*		-56	2.3(0.2) ≈ 2.3(0.2)		0
	a	107(15) > 63(13)*		-41	3.3(0.5) ≈ 3.4(0.6)		3
	d	112(26) > 61(14)*		-46	3.9(0.9) ≈ 4.0(1.4)		3
OH	w	125(19) > 57(14)*		-54	2.4(0.5) ≈ 2.5(0.5)		4
	m	149(31) > 58(21)*		-61	2.3(0.2) ≈ 2.5(0.8)		9
	a	104(10) > 56(15)*		-46	3.3(0.5) ≈ 3.0(0.4)		-9
	d	113(13) > 54(17)*		-52	3.9(0.9) ≈ 2.9(0.6)		-26
OO	w	125(22) > 60(15)*		-52	2.4(0.5) ≈ 2.4(0.3)		0
	m	145(49) > 56(14)*		-61	2.6(1.8) ≈ 2.4(1.1)		-8
	a	112(17) > 53(15)*		-53	2.9(0.5) ≈ 2.9(0.4)		0
	d	122(21) > 48(17)*		-61	3.1(1.3) ≈ 2.9(0.8)		-6

Maximum transmissibility at maximum acceleration spectral density (maxT@maxPSD), young and healthy (YH), older healthy (OH), older osteoporotic (OO), straight walking (w), walking and turning (m), stair descent (d), stair ascent (a), lumbar spine (l), thoracic spine (t), * = significant difference

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