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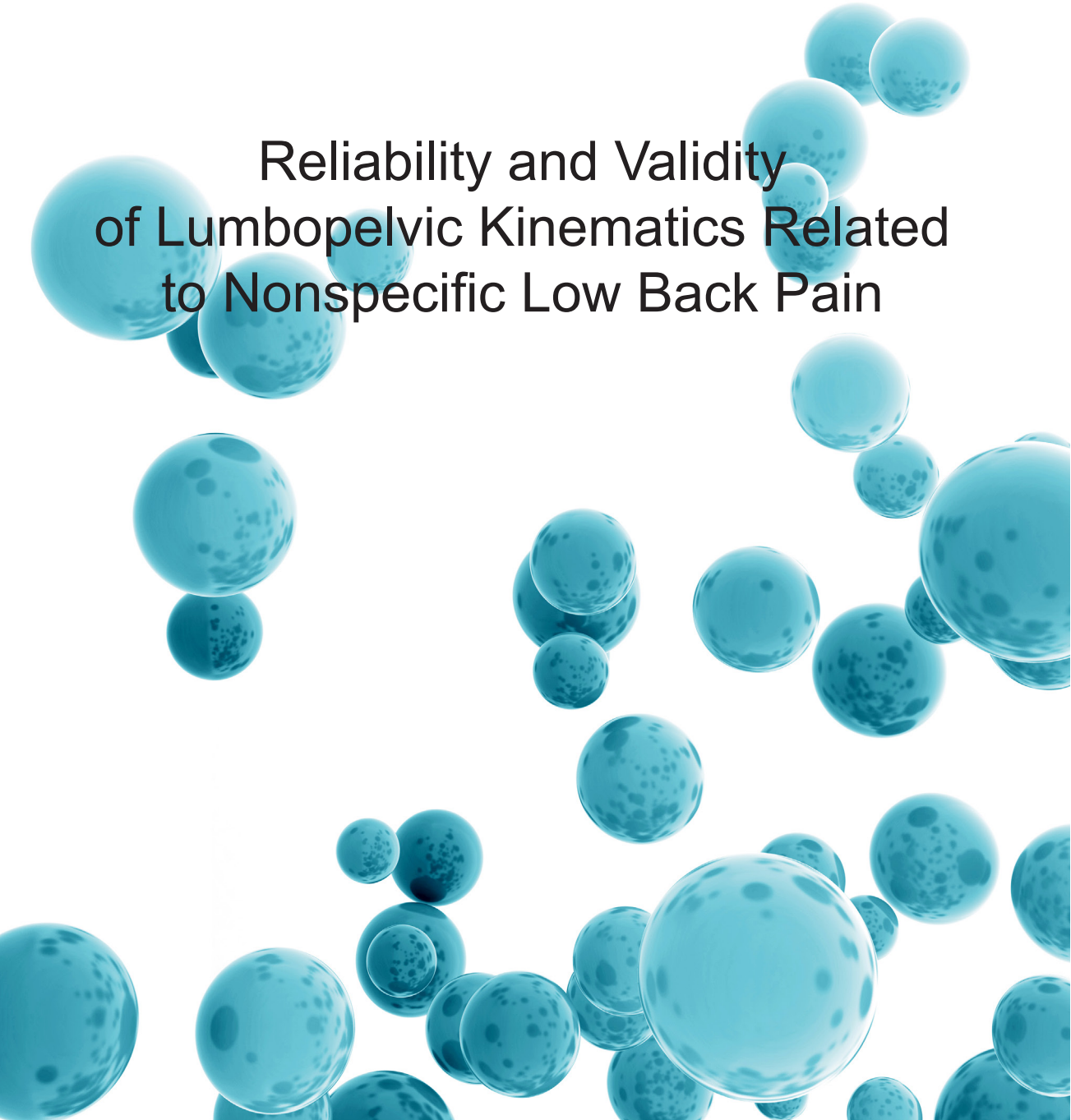
CHRISTOPH BAUER

Reliability and Validity of Lumbopelvic Kinematics Related to Nonspecific Low Back Pain



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ACADEMIC DISSERTATION
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UNIVERSITY OF TAMPERE

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"*a severe contest between intelligence, which presses forwards, and an unworthy, timid ignorance obstructing our progress.*"

The Economist, Mission Statement – attributed to Walter Bagehot

ABSTRACT

Low back pain (LBP) poses substantial challenges for clinical management. In 80-90% of all LBP patients, symptoms are attributed as nonspecific LBP (NSLBP). Attempts to identify effective interventions for people with NSLBP have often been unsuccessful. Past studies often treated NSLBP as a homogenous entity, although many clinicians notice distinguishable subgroups in daily practice. Valid classification systems are needed and are priority for primary care of LBP patients. A basic component of many contemporary LBP classification systems is the examination of lumbopelvic and postural kinematics. This examination is problematic because simple measurement systems such as visual observation or goniometers lack accuracy, reliability, validity, comprehensiveness, and practicality. To overcome these limitations this doctoral thesis introduces a novel, wireless movement-analysis system based on inertial measurement units (IMUs).

In using the novel IMU system, the aims of the studies conducted in this thesis were to assess: The concurrent validity of lumbopelvic kinematics (Study I), the reliability of lumbopelvic kinematics (Study II), the associations between NSLBP intensity and lumbopelvic kinematics (Study III), the associations between fatigue and NSLBP with non-linear lumbopelvic kinematics (Study IV), and the effect of exercise therapy on non-linear lumbopelvic kinematics (Study V).

Studies I-IV were conducted at a movement laboratory. Asymptomatic controls and subjects with nonspecific LBP performed a series of lumbar movement tests, from which indices of lumbopelvic kinematics (e.g. range of motion, movement variability and complexity) were calculated. The concurrent validity of the IMU-system was tested against an optoelectronic system. The reliability of lumbopelvic kinematics was analysed by comparing repeated measures over two days. To analyse the association between NSLBP intensity and lumbopelvic kinematics participants with different levels of NSLBP intensity performed movement tests. To investigate the effect of fatigue and NSLBP participants performed a movement test prior and after fatiguing of the lumbar musculature. The effect of exercise therapy was investigated in a randomized controlled trial in study V: After randomization, the intervention group was treated twice a week for six months while the control group only attended the

measurement sessions. Follow-up measurements were taken at post treatment and at twelve months follow-up.

The IMU system is concurrently valid to measure lumbopelvic kinematics in the primary movement direction. However, the system appears less valid for assessing movements in non-primary directions. On average, measures of lumbar range of motion, movement variability and complexity are more reliable compared to measures of movement control impairments and reposition error. NSLBP intensity affects lumbopelvic kinematics, so that participants with higher intensity NSLBP showed more variable and less predictable lumbar movement. Fatigue affects lumbopelvic kinematics, and this effect depends on the presence of NSLBP. The painfree participants showed more complex and less predictable lumbar movement after an isometric endurance test than participants suffering from LBP. Painfree people might adjust to fatigue by reducing load on fatigued tissues while preserving task performance. Exercise therapy affects lumbopelvic kinematics, and when compared to no intervention it may reverse or reduce deterioration of lumbar movement control, by increasing or preserving the degree of movement variability.

As a conclusion, this thesis identified concurrently valid and reliable indices of lumbopelvic kinematics related to nonspecific NSLBP. The association between 1) lumbopelvic kinematics and 2) NSLBP intensity, fatigue, and exercise therapy appears to be bidirectional: Painfree subjects show less variable lumbar movement than people with NSLBP, but they exhibit more complex and variable movement as a response to fatigue. Six months exercise therapy resulted in preserved, more variable and complex movement strategy. Therefore, a nonlinear or U-shaped relationship between movement complexity and variability with disease was identified.

Future research should address questions such as improvements of the IMU system's validity and define the optimal lumbar movement strategy that would be predictive for low back health in prevention and also in follow up of active physical rehabilitation.

TIIVISTELMÄ

Alaselkäkivun elinikäinen ilmaantuvuus on 80-85 %, joka luo merkittävän haasteen terveydenhuollolle ja yhteiskunnalle. Arvioiden mukaan 80-90 % selkäoireisista on epäspesifistä alaselkäkivusta johtuvaa, jossa nykytiedon mukaan katsotaan löytyvän useita alaryhmiä. Eri hoitomuotojen valinnoissa epäspesifin alaselkäkivun alaryhmien tunnistaminen on oleellisen tärkeää ja tähän soveltuvat uudet validit ja objektiiviset arviointimenetelmät mahdollistaisivat kohdennetumman ja tuloksiltaan paremman hoidon. Kliinisessä selkäpotilaan tutkimisessa lannerangan kinematiikan silmämääräiset arviot luovat perustan epäspesifin alaselkäkivun alaryhmien tunnistamiseen, johon kaivataan kipeästi objektiivisia valideja menetelmiä. Tässä väitöstyössä tutkittiin uuden langattoman inertiaalisen mittausmenetelmän (Inertial Measument Unit, IMU) ja siihen liittyvien lineaaristen ja epälineaaristen analyysimenetelmien validiteettia epäspesifin alaselkäkivun liikekontrollihäiriön tunnistamisessa ja kuntoutuksen edistymisen seurannassa.

Tämän väitöstutkimuksen tarkoituksena oli selvittää inertiaalisella mittausmenetelmällä arvioidun lannerangan kinematiikan (lineaariset ja epälineaariset parametrit) samanaikaisvaliditeettia (Tutkimus I), reliabiliteettia (Tutkimus II), yhteyttä alaselkäkivun intensiteettiin (Tutkimus III), yhteyttä lannerangan lihasten väsymiseen alaselkäkipuisilla ja terveillä (Tutkimus IV), ja validiteettia arvioimaan aktiivisen liikunnallisen kuntoutuksen aikaansaamia lannerangan kinematiikan muutoksia alaselkäkipupotilailla (Tutkimus V).

Tutkimukset I-IV toteutettiin laboratorio olosuhteissa. Terveet ja alaselkäkipuiset koehenkilöt suorittivat selän liikekontrollihäiriöiden tunnistamiseen käytettyjä liikesarjoja. Inertiaalisella mittauslaitteella lineaarisia ja epälineaarisia parametrejä käyttäen arvioitiin objektiivisesti lannerangan kinematiikkaa. Samanaikaisvaliditeetti määritettiin vertaamalla inertiaalisen ja optoelektrisen mittausmenetelmien tuloksia keskenään. Inertiaalisen mittausmenetelmän toistettavuus määritettiin vertaamalla perättäisinä päivinä saatuja tuloksia keskenään. Kivun intensiteetin ja lannerangan kinematiikan assosiaatio määritettiin alaselkäkipuisilla koehenkilöillä. Selkälihasten väsyvyyden vaikutus lannerangan kinematiikkaan määritettiin alaselkäkipuisilla koehenkilöillä vertaamalla mittaustuloksia ennen ja jälkeen maksimaalista vakioitua selän

lihasväsytystestiä. Aktiivisen liikunnallisen kuntoutuksen vaikutuksia uudella inertiaalisella mittausmenetelmällä arvioituun lannerangan kinematiikkaan tutkittiin satunnaistetussa kontrolloidussa tutkimuksessa. Alaselkäkipupotilaiden satunnaistamisen jälkeen aktiivinen liikunnallinen kuntoutusryhmä harjoitteli kahdesti viikossa puolen vuoden ajan ja kontrolliryhmä osallistui ainoastaan mittauksiin. Lannerangan kinematiikan mittaukset suoritettiin kuntoutusjaksoa edeltävästi ja jakson jälkeen, sekä vuoden seurannassa.

Tulosten perusteella inertiaalinen mittausmenetelmä todettiin validiksi menetelmäksi mittaamaan lannerangan kinematiikkaa primaarissa liikesuunnassa, mutta heikommaksi muissa liikesuunnissa. Inertiaalisella mittausmenetelmällä määritetty rangan liikelaajuus, liikkeen monimuotoisuus ja liikkeen kompleksisuus ovat toistettavampia kuin silmämääräisesti arvioidut liikekontrollihäiriön toteamiseen käytetyt menetelmät. Alaselkäkivun intensiteetin lisääntyminen todettiin johtavan monimuotoisempaan ja vähemmän ennustettavaan lannerangan liikemalliin. Alaselän lihasten väsyminen johtaa erilaiseen liikemalliin terveillä verrokeilla kuin alaselkäkipuisilla. Terveillä verrokeilla lihasten väsyminen johtaa kompleksisempaan ja vähemmän ennustettavaan lannerangan liikemalliin, joka on rangan rakenteiden kuormitusta vähentävä ja suojaava liikestrategia. Inertiaalista mittausmenetelmää käyttäen ja liikkeen monimuotoisuutta arvioiden, kuuden kuukauden aktiivinen liikunnallinen kuntoutus vähentää lumbopelvistä liikekontrollihäiriötä. Samaa ilmiötä ei havaittu passiivisessa kontrolliryhmässä.

Johtopäätöksenä uusi inertiaalinen mittausmenetelmä todettiin validiksi ja toistettavaksi mittausmenetelmäksi arvioimaan lannerangan kinematiikkaa epäspesifistä alaselkäkivusta kärsivillä koehenkilöillä. Selän kivun intensiteetin, selän lihasväsymisen ja liikunnallisen kuntoutuksen aikaansaamien muutosten yhteys lannerangan kinematiikkaan näyttäisi olevan kaksisuuntainen. Kivuttomilla koehenkilöillä rangan liikkeet ovat monotonisempia, mutta muuttuvat lihasväsymisen seurauksena kompleksisemmiksi kuin alaselkäkipusilla. Alaselkäkipuisten kuuden viikon liikunnallinen kuntoutus johtaa monimuotoisemman, kompleksisemman ja rangan rakenteita vähemmän kuormittavan liikestrategian lisääntymiseen. Näin ollen lannerangan liikkeen vaihtelevuus ja kompleksisuus assosioituu epälineaarisesti tai U-käyrän muotoisesti selän terveyteen.

Jatkotutkimuksissa tulisi keskittyä inertiaalisen liikemittausmenetelmän validiteetin parantamiseen. Tavoitteellista olisi tunnistaa ja löytää optimaalinen rangan liikemalli, joka olisi prediktiivinen selän terveyteen liittyen niin preventiossa kuin kuntoutuksen seurannassa.

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LIST OF ORIGINAL PUBLICATIONS

I. Bauer CM, Rast FM, Ernst MJ, Kool J, Oetiker S, Rissanen SM, Suni JH, Kankaanpää MJ. Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement. Journal of Electromyography and Kinesiology. 2015;25(5):782-90.

II. Bauer CM*, Heimgartner M*, Rast FM, Ernst MJ, Oetiker S, Kool J. Reliability of lumbar movement dysfunction tests for chronic low back pain patients. Manual Therapy. 2016;24:81-4.

III. Bauer CM, Rast FM, Ernst MJ, Oetiker S, Meichtry A, Kool J, Suni JH, Kankaanpää MJ. Pain intensity attenuates movement control of the lumbar spine in low back pain. Journal of Electromyography and Kinesiology. 2015;25(6):919-27.

IV. Bauer CM, Rast FM, Ernst MJ, Meichtry A, Kool J, Rissanen SM, Suni JH, Kankaanpää MJ. The effect of muscle fatigue and low back pain on lumbar movement variability and complexity. Journal of Electromyography and Kinesiology. 2017;33:94-102

V. Bauer CM, Kankaanpää MJ, Rast FM, Meichtry A, Rissanen SM, Suni JH. Efficacy of six months neuromuscular exercise on lumbar movement variability – A randomized controlled trial. In Preparation

* contributed equally

ABBREVATIONS

BMI	Body Mass Index
CNS	Central Nervous System
CT	Computerized Tomography
CTD	Cumulative Trauma Disorder
EMG	Electromyography
EZ	Elastic Zone
IMU	Inertial Measurement Unit
LBP	Low Back Pain
MAR	Missing at Random
MCI	Movement Control Impairment
MCMC	Monte-Carlo-Markov-Chain
MRI	Magneto Resonance Imaging
NME	Neuromuscular Exercises
NRS	Numeric Rating Scale
NSLBP	Nonspecific Low Back Pain
NZ	Neutral Zone
ODI	Oswestry Disability Index
OPT	Optoelectronic System
PS	Physical Stress
RE	Reposition Error
RM	Repeated Movement
RQA	Recurrent Quantification Analysis
VAS	Visual Analog Scale

NOTATIONS

95%CI	95% confidence interval
95%HPDI	95% highest posterior density interval
a_{fin}	Final angle
a_{max}	Maximal angle
a_{min}	Minimal angle
a_{start}	Starting angle
σ	Standard deviation of the underlying time series
Φ	Coefficient of determination
φ	Probability that two embedding vectors are similar
CE	Constant error
CV	Coefficient of variation
DET	Determinism
E	Expected error
L	Length of diagonal line
l_{max}	Maximal length of diagonal line
l_{min}	Minimal length of diagonal line
P(l)	Number of diagonal lines of length *l*
$R_{i,j}$	Recurrent point
r	Tolerance
r^2	R squared
REC	Recurrence rate
rmse	Root mean squared error
rmsj	Root mean squared jerk
ROM	Range of motion
RP	Recurrence plot
SaEn	Sample entropy

1 INTRODUCTION

Low back pain (LBP) is a major international health problem, and its lifetime prevalence of 80–85% poses substantial challenges for clinical management [1]. In the majority of patients with LBP (80–90% of all patients), symptoms are attributed to nonspecific LBP (NSLBP) [2]. These patients frequently seek conservative treatment from rehabilitation specialists such as doctors and physiotherapists. A minority of patients present clinical characteristics, called red flags, which indicate more serious and specific causes of LBP. Such patients need referral to appropriate medical practitioners, and they must undergo specific medical screening, including imagining techniques, to identify the underlying medical condition and guide therapy. Attempts to identify effective interventions for people with NSLBP have often been unsuccessful [3]. Many past studies have treated it as a homogenous entity, although many clinicians notice distinguishable subgroups in daily practice. The development of valid classification systems is therefore a priority for the primary care of patients with LBP [4, 5]. The literature describes a variety of systems [2, 6-9] whose underlying premise is that classifying patients into groups based on clinical characteristics and matching these subgroups to tailored interventions will improve the outcome of therapy [3]. A basic component of many contemporary classification systems is the examination of lumbopelvic and postural kinematics [7, 9-12]. This examination is problematic because simple measurement systems such as visual observation or goniometers lack accuracy, reliability, validity, comprehensiveness, and practicality [13, 14]. Optoelectronic measurement systems are reference standards for noninvasive analyses of lumbopelvic kinematics within research settings [15, 16]. They are not applicable in daily clinical practice, however, due to their high cost, required installation space, specific marker placement, and the subsequent data capture, analysis, and processing necessary. These factors limit the analysis to several standard procedures [17]. To overcome these limitations, this doctoral thesis introduces a novel, wireless movement-analysis system based on inertial measurement units (IMUs), and it focuses on the concurrent validity of this system, as well as the reliability of the lumbopelvic kinematics it measures. Furthermore, this thesis explores the effects of NSLBP intensity, fatigue, and exercise therapy on lumbopelvic kinematics.

2 LITERATURE REVIEW

2.1 Low Back Pain

2.1.1 Epidemiology

LBP is a major international health problem, with a lifetime prevalence of 80–85%, and it poses substantial challenges for clinical management [1]. Best estimates suggest that the prevalence of chronic LBP is about 23%, disabling 11–12% of the population [18-20]. The 1-year incidence of a first episode ranges from 6.3 to 15.3%, while estimates of the 1-year incidence of any LBP episode range from 1.5 to 36% [19, 21]]. Adult incidence of chronic LBP is estimated at 5% per year [22]. The costs of LBP to society are staggering. In 2005, health care expenditures resulting from this condition were estimated at €2.6 billion in Switzerland, which corresponds to 6.1% of the total health care expenditure (€33.6 billion) or 2.3% of the gross domestic product. These values do not take into account annual lost productivity due to LBP, estimated as at least €4.1 billion [23]. Throughout the world, it is, amongst diseases such as osteoarthritis, a leading cause of activity limitations and work absences, and it is associated with economic burdens [24-27]. Between 24% and 33% of people who have experienced activity-limiting LBP had reoccurring episodes [28, 29].

LBP's prevalence varies, based on factors such as sex, age, education, and occupation [3]. The two major categories of suspected risk factors for first episodes are individual and activity-related factors involving both work and leisure [3]. Individual factors can be genetic, demographic, anthropometric, physical, psychosocial, and more. Genetic factors link to specific disorders of the spine such as disc degeneration [30], but their links to LBP remain questionable [3]. Despite the relationship between genetics, anthropometrics, and early environmental influences with degenerative changes of the spine, these factors appear to be only weakly related to LBP [3, 31-33]. Women have a higher prevalence of LBP than men [34-37]. Increased age brings a higher prevalence and more severe forms as well [38], and this association peaks between the ages of 60 and 65 [39, 40]. Lower educational status is associated with increased prevalence, longer episodes, and poorer prognosis [3, 19, 38, 41, 42]. Higher occupational physical

demands are associated with higher LBP prevalence: for material workers it is 39%, whereas for sedentary workers it is 18.3% [19, 43]. LBP is associated with operating heavy equipment and transferring heavy loads [44], but the evidence is inconclusive for the relationship between risk of LBP and trunk muscle strength or spinal mobility [45].

Psychosocial factors such as fear and depression seem to play a larger prognostic role than physical factors [3]. Avoidance behaviour resulting from fear is important when pain has become persistent [46, 47], while distress and depression play important roles in the early stages of LBP [48]. Changes in behaviours and reductions of disability may be more important for successful treatment of chronic LBP than physical factors [49].

Although some individual and lifestyle factors put one at risk for LBP, they may not be associated with recovery from it. Duration of sick leave due to LBP is not influenced by a previous history of the condition, job satisfaction, educational level, marital status, number of dependents, smoking, working more than 8-hour shifts, occupation, or size of industry or company [3, 25]. In addition, the clinical course for patients with comorbidities is just as favourable as for those without [3, 50]. Other factors may be more important: One's own expectation of recovery appears to be an important predictor for the decision to return to work, since people with higher expectations have fewer sickness-related absences [3, 50]. Furthermore, LBP intensity, high job satisfaction, and an active coping style are predictors for a better outcome following an episode [51].

In summary, no definitive cause for initial episodes of LBP is known. Risk factors are multifactorial, population specific, and associated only weakly with its development [3].

2.1.2 Natural History

Clinical practice guidelines claim that most individuals with acute LBP recover within 4–6 weeks, while those with chronic LBP have a poorer prognosis [52, 53]. Recent research indicates, however, that the natural history of LBP varies [28, 54-56]. Patients with acute and chronic LBP improve markedly in the first six weeks after inception, for both pain and disability [56]. Improvement slows after that, and low to moderate levels of pain and disability can still be present after a year, especially in patients with chronic pain [56]. A study of patients visiting a primary-care clinic found that 72% of patients with acute LBP and 42% of those with chronic LBP had completely recovered by 12 months [54, 57]. These data suggest that the good prognosis for patients with acute LBP

is overestimated and that the potential for improvement in people with chronic LBP is underestimated [55].

Between 33 and 82% of people with LBP report one or more recurrences within one year [28, 54, 57-59]. Recurrent episodes change the prognosis for recovery from an LBP episode: it becomes less favourable and more variable [59]. The unfavourable prognosis of recurrent LBP prioritises research on preventive interventions.

2.1.3 Pathoanatomical Features

Any innervated structure in the lumbar spine can cause low back and referred pain [3]. These structures include those of the three subsystems that preserve spinal stability and function: the passive, active, and motor control subsystems (see Section 2.2) [60, 61]. Disc degeneration is claimed to be the main cause of LBP in adult patients (42%), followed by zygapophysial joint pain (31%) and sacroiliac joint pain (18%) [62]. One might expect advances in imaging techniques to have increased the likelihood of detecting a link between pathology and pain in the lumbar spine, yet false-positive findings (e.g., people without LBP who have abnormal findings) make determining the pathoanatomical origins of LBP difficult [3].

For example, computerized tomography (CT) scans, magnetic resonance imaging (MRI), and myelography show herniated discs in 20 to 76% of people without sciatica type LBP [31-33]. Furthermore, 53% of people with LBP and 32% of people without LBP presented abnormal findings such as disc degeneration or nerve root compression. Longitudinal studies using MRI report that LBP can develop in the absence of abnormal findings regarding the lumbar spine [63]. Finally, as further evidence for the overrating of imaging methods, researchers monitored people for five years who had a herniated disc and no LBP. They discovered that physical and psychological work factors were more powerful than imaging techniques in predicting the need for an LBP-related medical consultation [3, 64].

In summary, the association between clinical complaints and abnormal imaging findings must be considered with caution. Even when abnormalities are present, establishing direct cause and effect between the pathological finding and the patient's condition is elusive [3]. This discrepancy might not assist in patient management.

2.1.4 Classification of Low Back Pain

LBP can be divided into acute, subacute, and chronic phases. While different definitions are reported, those commonly accepted are less than 1 month since the onset of one LBP episode, between 2 and 3 months, and longer than 3 months [3]. The exclusive use of temporal criteria to describe the clinical course of LBP is insufficient, however, since the time until improvement from a single episode does not take into account recurrent ones [3, 65, 66].

In the majority of patients with the condition (80–90% of all patients), symptoms are attributed to NSLBP [2]. A minority of patients present clinical characteristics, called red flags, indicating specific, more serious causes. Such patients require referrals to appropriate medical practitioners. They must also undergo specific medical screening, including imagining, to identify the underlying medical condition. Relevant conditions include tumours, compression fractures, and abdominal aortic aneurysms. In addition, psychosocial and social factors (called yellow flags) may contribute to a patient's transition from acute to chronic LBP and prolonged disability [3, 32]. Attempts to identify effective interventions for people with NSLBP have often been unsuccessful [3]. Past studies often treated it as a homogenous entity, although many clinicians notice distinguishable subgroups in daily practice. The development of valid classification systems is a priority for the primary care of patients with NSLBP [4, 5].

The literature describes a variety of classification systems [2, 6-9]. Their underlying premise is that classifying patients into groups based on clinical characteristics and matching these subgroups to tailored interventions will improve the outcome of physiotherapy [3]. Since subgrouping based on pathoanatomy is limited due to false-positive findings, emphasis is placed on patterns of clinical signs and symptoms to identify pain-driving factors in individual patients [2, 3, 6-9, 67-69]. These classification systems often focus on excluding red flags, recognizing yellow flags, identifying pain-driving factors, and tailoring nonsurgical interventions towards the individual patient's needs [2, 3, 6-9, 67-77].

One example with established reliability and validity is the O'Sullivan classification system, which differentiates between different classes of specific LBP and NSLBP. The first stage involves screening for potential red flag disorders and determining whether the condition is specific or NSLBP. The second stage considers whether NSLBP disorders have an adaptive or maladaptive response to the disorder, which manifest in centrally or peripherally mediated NSLBP. Centrally mediated pain is further subclassified into the presence of nondominant or dominant yellow flags. Peripherally mediated disorders are subclassified into either NSLBP or a pelvic girdle pain disorders.

Peripherally mediated lumbar spine pain disorders are divided into movement impairment or movement control impairment (MCI) disorders. If the lumbar spine is the source of pain, the primary directional provocation bias as well as the symptomatic spinal level are noted (Figure 1) [2, 72, 78-80].

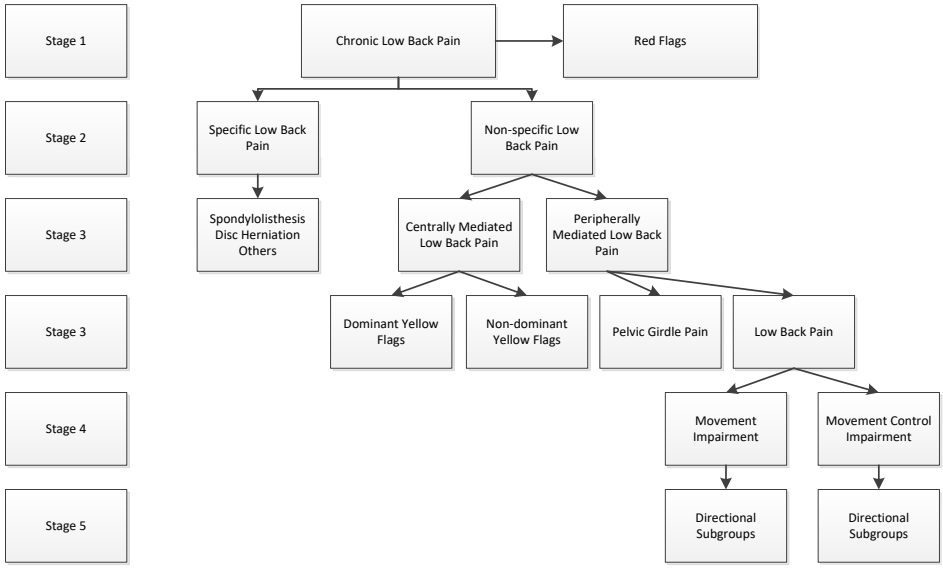

Figure 1. Extract from the O'Sullivan classification system; Adapted from O'Sullivan (2005) and Vibe-Fersum and colleagues, 2009 [2, 79].

2.1.5 Examination and Differential Diagnosis

The goals when examining patients with LBP are to determine 1) the presence of red and yellow flags, 2) the presence of clinical findings that indicate an impairment of body functions, body structures, and limitations in activity and participation, and 3) changes in impairments and limitations after treatment [3]. Clinicians should choose the most relevant outcome measures based on the patient's presentation, needs, and goals. In particular, they should screen for and address prognostic factors for recurrent and chronic LBP. Consensus documents agree on a "core" set of domains that should be covered in outcome assessment of LBP, including generic health status, pain, back-specific function and disability, work disability, physical impairments, psychosocial factors, and patient satisfaction [3, 81, 82].

2.1.6 Treatment Options

A large body of research describes multiple interventions for the treatment of LBP, which primarily target impairments of body structure, function, and limitations in activity and participation in people suffering from NSLBP. Generally, an early physical rehabilitation or physiotherapy intervention reduces the risk of a transition from acute to subacute or chronic NSLBP [83-90]. The European guidelines for the management of chronic NSLBP recommend exercise therapy as a first-line treatment, [4, 91] as high-quality evidence suggests that it is more effective than other interventions for this condition [92]. In contrast, high-quality evidence on the effect of other interventions is often lacking [93-104]. The quality of evidence ranges from high to very low that motor control exercises, spinal manipulation therapies, muscle energy techniques, therapeutic massage, therapeutic ultrasound, traction, or back schools provide similar outcomes to other forms of interventions or, in some cases, to sham or no interventions [93, 94, 100-103, 105].

Neuromuscular exercise (NME) such as Pilates are administered to improve movement control, flexibility, and strength [104]. The Pilates method is currently one of the most popular exercise programs used in clinical practice [104]. Low to moderate quality evidence indicates that NME is 1) more effective than minimal interventions for pain and disability and 2) more effective than other exercises for function, at an intermediate follow up, for patients with acute, subacute, and chronic NSLBP [106-115]. However, there is currently no evidence if NME induces changes in lumbar movement control.

Movement-control exercises focus on restoring motor control and on improving and correcting pain-provoking movement patterns and postures. Low-quality evidence indicates that movement-control exercises are more effective than other interventions in improving NSLBP intensity and disability in people with movement-control impairments, at least in the short term, but the effect sizes of the respective studies vary [98]. This indicates that thorough multilevel classification of patients with LBP is mandatory to assure that treatments match the individual patients' presentation. As compared to other interventions, the existing, low-quality evidence conflicts as to whether movement-control exercises are more effective for NSLBP intensity and disability in the long term [98]. The effect of both NME and movement-control exercises might be increased by customizing the intervention to specific movement-control problems, such as with flexion or active extension [2]. However, customization requires reliable movement-control tests. Unfortunately, data on the reliability of such observation-based tests are discouraging [13]. Objective, quantitative measures of

movement control, as provided by systems using IMUs, can overcome those limitations. Furthermore, quantitative, interval-scaled measures provide detailed insights into the relations between NSLBP, movement control, and exercises.

2.2 Functional Anatomy of the Lumbar Spine

2.2.1 Mechanical Functions of the Lumbar Spine

Spinal functionality is the capacity of the spinal column to remain coherent and preserve its normal functionality during physiological postures and movements [116-118]. The term indicates that the main function of the musculoskeletal system of the spine, besides protecting the spinal cord, is providing passive and active stability during static postures, trunk movement, and locomotion.

A functional unit of the spine consists of two adjunct vertebrae and the interconnected intervertebral disc. Each unit can perform three translational and rotational movements along the x, y, and z Cartesian axes of space. The respective axis denotes translational movements. Rotational movements in the sagittal, frontal, and transverse plane are called flexion-extension, lateral flexion, and axial rotation. Imaging techniques such as stereo radiography quantify translational and rotational movements between individual vertebrae and the entire lumbar spine. Noninvasive techniques such as IMU systems are cost-efficient, low-risk alternatives for quantifying lumbopelvic kinematics on the body's surface.

2.2.2 Stability: The Passive, Active, and Motor Control Subsystems

Spinal stability is preserved by three interconnected systems that maintain the spine within physiological limits so as to avoid constant deformity, neurological deficits, or pain [61]. These subsystems consist of 1) the spinal column and ligaments, or passive subsystem, 2) the muscles and tendons, or active subsystem, and 3) the central nervous system (CNS) as a motor control [118, 119]. According to this principle, the passive subsystem controls the elastic zone (EZ) while the active subsystem and the CNS control the neutral zone (NZ) [118]. The NZ is the initial part of intervertebral motion on either side of the neutral position [118]. Motion in the NZ meets relatively low resistance. The subsystems exhibit high flexibility due to the laxity of joint capsules, ligaments, and tendons (Figure 2) [61, 118]. The EZ follows the NZ. Motion in the EZ

meets significant resistance due to increased stiffness [61, 118]. The resistance to movement and the slope of the load displacement curve increase linearly when the ligaments, capsules, fasciae, and tendons are subjected to tension (Figure 2) [61, 118]. Each lumbar vertebra has its individual range of motion (ROM), NZ, and EZ, in all six degrees of freedom [119]. In asymptomatic subjects, the NZ and ROM are contained within the limits of the painfree zone [118, 120].

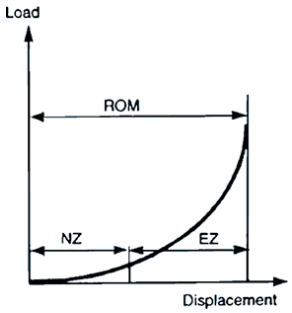

Figure 2. Load displacement curve. EZ = elastic zone; NZ = neutral zone; ROM = range of motion. The load/displacement curve of the spine is not linear. The ROM of the spinal joints includes the initial NZ, with relatively large displacements at low load and EZ requiring more loads per unit of displacement because of the tension of capsules and ligaments. Taken with permission from Adams and colleagues, 2006 [116].

Passive Subsystem

The passive subsystem of the lumbar spine includes the five lumbar vertebrae, discs, disc–intervertebral joints, and ligaments (Figure 3 and Figure **4**). The lumbar vertebral bodies resist most of the vertical forces acting on the spine [116], which arise from gravity and the tension of the paraspinal muscles [116, 117]. The vertebral bodies' resistance depends on their bone mineral density, in particular the dense trabeculae network between their endplates [116]. Consisting of vertical, horizontal, and two oblique subsystems, the trabecular network disperses the vertical forces laterally. The perforated vertebral endplates consist of thin cortical bone, so they absorb vertical forces minimally [121]. The vertebral arches, composed of the pedicles and lamina, form the foramen and serve as attachments for muscles and ligaments; thus, they absorb and transmit forces from the active stabilization subsystem. The zygapophysial joints have several functions: 1) they stabilise the lumbar spine under compression and prevent excessive rotation and translation of individual vertebra during loading [116]; 2) they can transmit up to 70% of spinal compressive forces between vertebrae [122]. Finally, due to their shape they resist forces by acting perpendicularly to their articular surfaces,

limiting axial rotation of the lumbar spine and resisting forward shearing forces [123-127]. Their stabilising action can be enhanced by downward pulling of the lumbar spinous processes by the active stabilization subsystem [128].

The stabilizing action of a ligament depends not only on its intrinsic strength, but to a greater extent on the length of the lever arm through which it acts, the distance between the bony insertions, the point of force application, and the instantaneous axis of rotation of the vertebral body [118]. A strong ligament with a short lever arm may contribute less to stability than a weak ligament working by a longer lever arm, which provides a mechanical advantage [118]. The interspinous, supraspinous, zygapophysial joint capsular, intertransverse, and anterior longitudinal ligaments provide stability or resistance to movement. The main actions of the ligamentum flavum and the posterior longitudinal ligaments are to line the intervertebral foramen and protect the spinal cord from herniated disc material, respectively [116, 118, 122, 129-134].

The intervertebral discs' primary function is to transfer compressive forces between vertebrae, while allowing small intervertebral movements. They have the compression-resisting properties of joint cartilage and the tension-resisting properties of ligaments. An intervertebral disc allows and controls vertical compression and distraction, flexion–extension, lateral bending, and axial rotation, but is less capable of absorbing vertical forces or resisting movement [116, 118]. The intact nucleus pulposus and the encircling lamellae of the anulus fibrosus transfer vertical forces equally across the disc to the adjacent vertebrae. The anulus fibrosus consists of several layers of laminae, with different characteristics and functions. The outer lamellae function as ligaments and resist translation and rotation of adjacent vertebrae, due to their high content of type 1 collagen fibres [135]. The middle and inner lamellae are deformable and behave fluidly, accepting the tensile stress from the nucleus under vertical loading [136].

The cartilage endplates, which consist of hyaline cartilage, cover the central region of the vertebra's endplates and help to equalise the distribution of the vertical forces on the vertebra body and protect it from migration of the nucleus pulposus into the pores of the vertical body [116, 137].

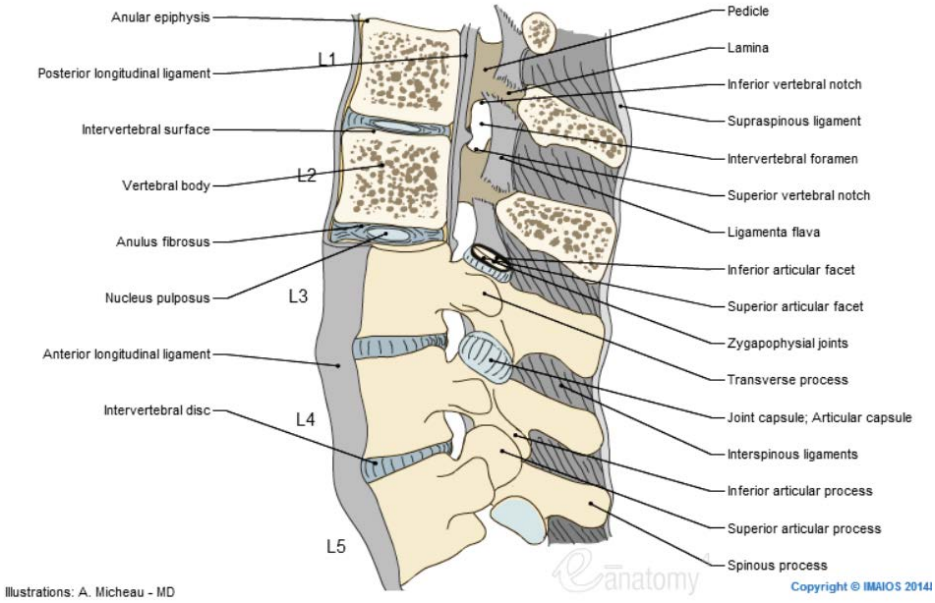

Figure 3. Anatomy of the lumbar spine, lateral view. From Imaios.com, with permission.

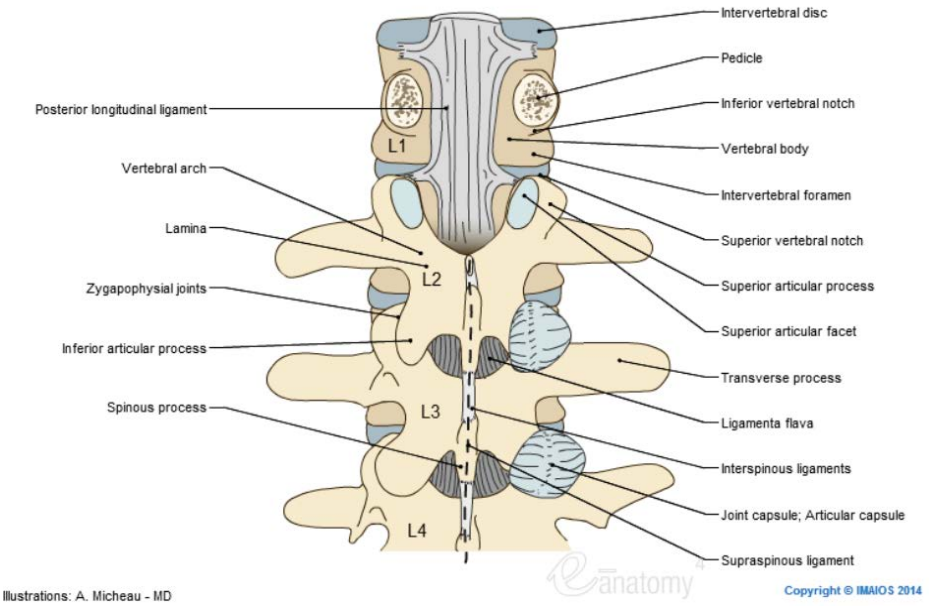

Figure 4. Anatomy of the lumbar spine, posterior view. From Imaios.com, with permission.

Active Subsystem

The spinal muscles are divided into three groups: superficial, intermediate, and deep muscles (Figure 5) [138-140]. The superficial and intermediate groups are extrinsic muscles that produce both respiration and translational/rotational movements of the vertebral column [141]. The synergistic and antagonistic action of groups of superficial and intermediate muscles actuates and controls translational and rotational movements of the lumbar spine. Concentric, bilateral contractions of the rectus abdominus, external and internal oblique, and psoas minor muscles produce flexion of the lumbar vertebra. Concentric contraction of the erector spinae, interspinales, quadratus lumborum, and multifidus muscles of both sides produces extension. The hip extensor muscles gluteus maximus and biceps femoris are synergists during lumbar extension throughout specific movements [141, 142]. Producing lateral flexion are the quadratus lumborum, intertransversarii, external and internal oblique, rectus abdominis, erector spinae, and multifidus. Concentric contraction on the contralateral side of the multifidus, rotatores, semispinalis, and external and internal oblique produces axial rotation of the trunk to one side.

The length of the lumbar and abdominal muscles, the angles at which they pull on veretebrae, and the lever arms are affected by posture [116]. Length might be the most important muscle property affected by posture, since it affects the maximum active tension [116]. Produced by the contractile elements, active tension might be greatest between 100 and 110% of a muscle's resting length, and it is diminished in a shortened or stretched muscle. Passive tension, which is produced by collagenous tissue, increases with muscle length [116]. Overall tension in the muscle is greatest when it is fully stretched [116]**.** Therefore, the back muscles do not produce their maximum active tension in a neutral lumbar lordotic positon. Accordingly, lumbar muscles are stronger when flexed, compared to a lordotic, neutral, or fully flexed position [143, 144]. The passive tension produced by the lumbar muscles becomes more important for lumbar stability with increased lumbar flexion, as the back muscles receive support from the lumbodorsal fascia and intervertebral ligaments [116, 117]. The erector spinae pull the lower lumbar vertebrae backward during upright standing in a neutral posture. This action limits the gravitational shearing force acting on them. The pulling angle is increased in a lordotic posture compared to a flexed one [145]. The lever arms of the lumbar muscles relative to the flexion–extension axis of rotation are influenced only marginally by different postures [146]. Co-contractions of the abdominal muscles increase stability of the lumbar spine and axial compression in the L4–L5 region [147].

Individual differences in recruitment patterns, compressive loading tolerances, and stability demands guide clinical decisions regarding exercise design and prescription during the rehabilitation of patients with NSLBP [147], who often require low-loading, high-stability exercises [148].

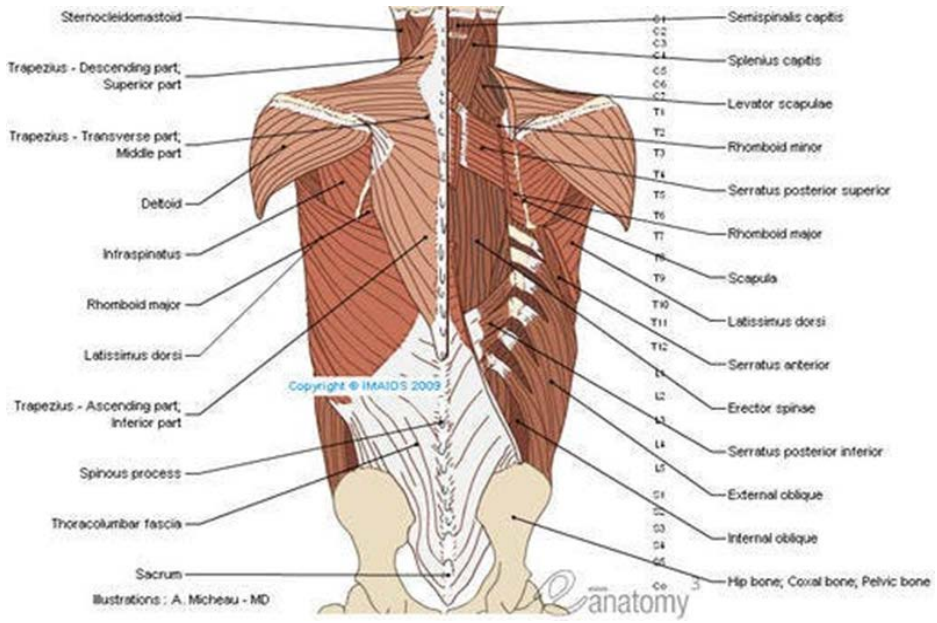

Figure 5. Muscles of the trunk, posterior view. From Imaios.com, with permission.

Motor Control System

Motor control is defined as the ability to regulate or direct physical movements and posture by activating and coordinating muscles [149]. Motor control is organized hierarchically, and final commands in the brain are issued based on the integration of feedback and feedforward information [149]. Feedback is provided by multisensory information from the periphery but also by the current state of the body when determining the appropriate set of muscle forces and joint activations to use in generating the desired movement [150]. Motor cortices provide feedforward information that is perceived as an efference copy of the induced motor action intended for somatosensory brain regions. This internal information exchange can then be compared with the peripheral feedback from sensory information of actual movements and can enable the brain to predict the effects of an action and to correct motor output accordingly [150].

Receptors in the periphery deliver afferent multisensory information. These receptors are positioned in the skin (pacini corpulses, merkles discs, meissners corpuscles, or ruffini endings), muscles (muscle spindels), tendons (golgi tendon organs), and joint capsules (fibrous capsules) [151]. Sensory transduction from these receptors enters the CNS at the dorsal horn of the spinal cord. At the segmental level, they are important during segmental reflexes; for example, in reacting to an acute injury [152]. The signals then ascend through the spinal pathways to the thalamus. The ascending anterolateral and dorsal column–medial lemniscal systems convey information to the brain on crude touch, pressure, pain, temperature, and proprioception [153]. The ascending pathways convey to the primary and secondary somatosensory cortices (S1/S2) through the thalamus. These cortices establish communication, with different associated brain areas in the parietal lobes and the cerebellum processing the information required for adequate motor behaviour [153, 154].

Beside somatosensensory and associated cortices, other brain structures are important in motor control. With complex anatomical connections, the basal ganglia consist of a group of nuclei at the base of the cerebral cortex that include the putamen, caudate nucleus, globus pallidus, subthalamic nucleus, and substancia nigra [150, 151, 155]. Among other motor tasks, they coordinate the accuracy of motor responses during reciprocal movements and between muscles in different extremities. Basal ganglia diseases (such as Parkinsonian diseases) are hypo- or hyperkinetic in nature [151]. Furthermore, the cerebellum receives information from the sensorimotor cortex and the periphery [149]. By comparing central and peripheral input, the cerebellum coordinates and fine-tunes movements; therefore cerebellar disorders result in ataxia.

The topical representation of the somatosensory body surface is located in S1, while the primary motor cortex contains sensorimotor representation. A homunculus illustrating these representations shows the arrangement of body parts receiving afferent peripheral somatosensory information [151](Figure 6) (the size of a body area's representation in the brain varies with the density of peripheral sensors). As the sensorimotor system is plastic, it can be altered by habituation, training, and pain [156]. Examples of pain conditions that affect representation in S1 are chronic LBP, generalized pain, chronic regional pain syndrome, or phantom limb pain [157-163].

Based on multisensory integration, primary and premotor motor cortices plan and initiate motor behaviour. Motor cortex output descends via polysynaptic connections through the corticospinal tract into the ventral horn of the spinal cord, more specifically the motor neurons. They transduce signals, which activate the motor units of muscles and the motor response to the neuromuscular junction through peripheral efferent

nerve fibres. The neuromuscular junction forms the chemical contact between the motor neuron and the muscles involved in the motor response [151].

The aforementioned peripheral receptors provide feedback information from motor behaviour. These information ascents into the CNS for comparison/adaptation or termination of the motor response. This learning process of conditioning and habituation forms the basis of CNS activity. If a motor behaviour is performed repeatedly with sufficient intensity, the motor response will become automated [151]. In contrast, responses not needed or used infrequently will become weaker and diminish [151]. Mosely described this process as the "use it or lose it" principle of neurophysiology [160].

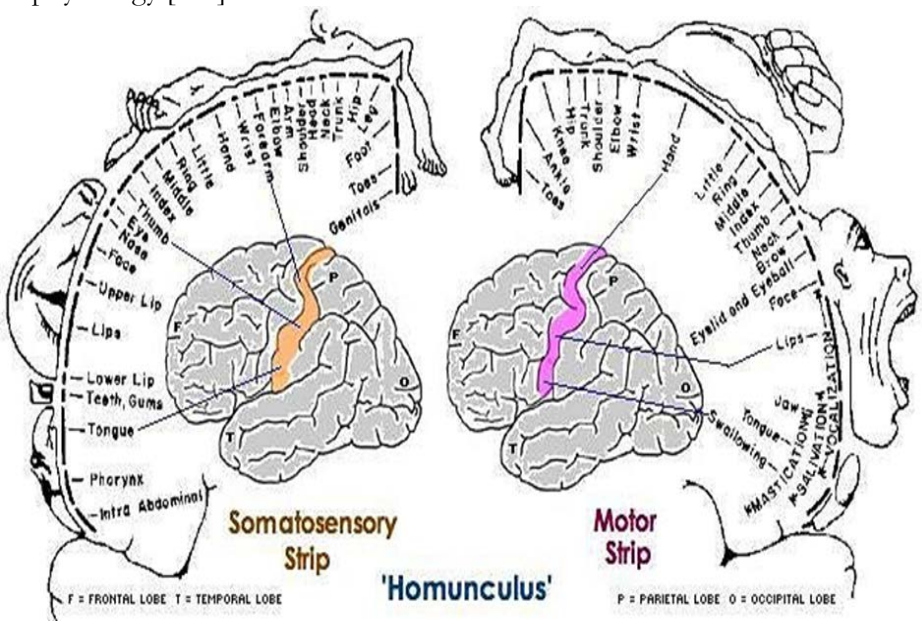

Figure 6. Homunculus of the somatosensory and motor cortex. With permission from pinterest.com

The balance control system regulates equilibrium and maintains the body's line of gravity within the base of support [164]. Balance control strategies are described as feedback mechanisms derived from the interaction of sensory input, sensorimotor integration, and adapted motor output [165]. Balance control requires input from multiple peripheral sensory systems, including the vestibular, somatosensory and visual systems [166]. The vestibular system transmits afferent directional information while relating the head position to gravitational acceleration [149]. The somatosensory system delivers afferent feedback from proprioception, pressure, vibration and kinaesthesia [167]. The visual system provides reference to the verticality of head and body motion

and spatial location relative to objects [149]. Multiple sensory inputs are combined with prior experience and converged to a set of movement possibilities [168]. From these possibilities, control priorities are selected and passed to the motor system, which generates coordinated inhibition and excitation of the entire active system [168, 169]. Within a perception-selection-motor feedback loop, the following two mechanisms work together. (1) Acting through central selection and optimization pathways (e.g., motor cortex), the slow intentional system allows sequential optimization, selection and inhibition of alternative possibilities of up to four selections per second [168, 170]. (2) Acting through previously facilitated transcortical, brain stem and spinal pathways, the fast habitual reflexive system implements coordinated responses to environmental stimuli with a latency as low as 50–100 ms [168, 171]. Several studies suggest that somatosensory dysfunctions — specifically proprioceptive ones — explain impaired motor control in people suffering from recurrent LBP [172-175]. In an attempt to compensate for proprioceptive deficits and to protect the painful area, such individuals may use trunk muscle activation strategies aimed at trunk stability [172, 176, 177]. Pain-avoiding compensations may potentially be a factor in the neurophysiological mechanisms of balance deficits associated with the recurrence of LBP [172, 178, 179].

2.3 Mechanical Loading and Low Back Pain

2.3.1 Biomechanical Aspects

High stress concentration at the intervertebral discs, ligaments, and zygapophysial joints can result from loading at unusual postures. Sustained creep (laxity) loading might diminish absorption of vertical forces by the intervertebral discs and narrowed vertebral arches. Furthermore, altered motor control, such as asymmetrical muscle activity, can induce abnormal, asymmetrical stress on the spine [116, 117]. Practical examples are occupations involving repetitive, heavy physical work (e.g., nursing), which may accelerate functional pathologies or cumulative trauma disorders (CTD) in the lumbar region [152, 180, 181]. Typically, CTDs in nurses are caused by repetitive lifting and transferring patients while maintaining ergonomically poor postures, such as working in a stooped position with a twisted back [182-186]. Cumulative load in the EZ induces height loss of the intervertebral discs; combined with insufficient rest time, it predicts a clinically relevant decline in low back function [187, 188]. Patient-handling activities that do not involve lifting (such as patient care) contribute to cumulative stress on lumbar

soft tissues, since they often require extreme spinal postures and comprise the majority of the work shift [189, 190]. Therefore, high-risk factors for CTD of the lumbar spine include high-magnitude loads, long loading durations, many repetitions, high movement velocities, and short rest periods in between work sessions [181, 188, 191]. Continued exposure to high-risk factors past the acute phase may lead to a chronic stage of CTD [181]. Cyclic tissue loading of the lumbar spine is associated with increased compressive and anterior/posterior shear forces at lumbar vertebra endplates, inducing increased cytokine expression levels and neutrophil density [192]. Prolonged cyclic work has been shown to induce transient creep/laxity in the spine, muscle spasms, and reduced stability, followed by acute inflammation/tissue degradation, muscular hyperexcitability, and increased local stability in the lumbar spine [180, 181, 193]. Figure 7 provides a schematic presentation of lumbar creep, muscular activity, cytokine expression, and stability during cyclic, repetitive, high-load loading at high velocities of the lumbar spine following rest [181, 194-204].

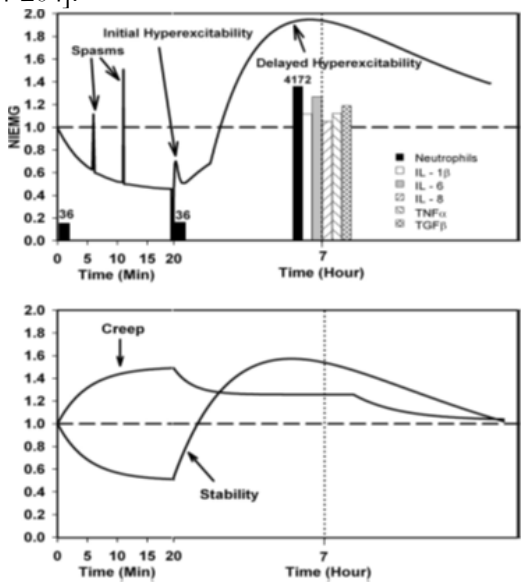

Figure 7. Comprehensive model of lumbar creep, neuromuscular control, tissue biology, and stability during and following high-load loading at high velocities (with permission from Elsevier Ltd. [181]).

2.3.2 Motor Control Aspects

Antagonistic activity of lumbar muscles (e.g., lumbar spine flexors and extensors) stabilises the lumbar spine by minimizing the need to recruit additional muscles during

unexpected events such as perturbations or sudden loading [189]. However, due to the compressive forces based on antagonistic muscle activity, it is theorized that an optimal level of antagonistic muscle activity exists, and that this level depends on the type of activity and posture [116]. Antagonistic muscle activity is high in situations where a high level of stability is required and is low during activities with inherently high compressive forces, such as lifting heavy objects [205-210]. Increased antagonistic activity has been described as "splinting" in people suffering from LBP [211]. Splinting is regarded as a protective co-contraction after an initial painful event that reduces the risk of further damage to the tissues involved [211]. The muscle tension associated with splinting imposes elevated compressive forces on the spine and can lead to prolonged pain and loss of function. It becomes maladaptive if it exceeds the normal period of inflammation and tissue repair [2, 212, 213]. Splinting is also associated with the diminished flexion–relaxation of antagonistic muscles observed in people suffering from chronic LBP [214]. To prevent injury, rapid spinal movements cause back muscles to contract and decelerate the body. These contractions may be reflexive in nature [215]. These reflexes diminish during prolonged or repeated stretching of the interspinous ligaments, reducing the ability of the back muscles to protect the spine by co-contraction of antagonists [213, 216]. This process can be explained partly by altered proprioception, such as desensitisation of stretch receptors through overuse or creep of ligaments [213, 216]. Other factors that diminish protective muscle activity are chronic joint pain and swelling, which inhibit local muscles via a short or long loop reflex [217, 218].

2.4 Kinematic Assessment of Lumbar Spine Function

2.4.1 Kinematic Assessment Methods

A basic component of examining patients with NSLBP is observing lumbopelvic and postural kinematics [7, 9-12]. A core component of many contemporary classification systems, this observation includes basic assessments of ROM, posture, proprioception, kinematics of activities of daily living, and complex movement-control tests selected depending on the chosen classification system and patients presentation [2, 6-9]. The examination of lumbopelvic kinematics is problematic in clinical settings because simple measurement systems such as visual observation, goniometers, tape measures, video cameras, or inclinometers lack accuracy, reliability, validity, comprehensiveness, and practicality [13, 14, 219]. Optoelectronic measurement systems are reference standards

for noninvasive analyses of lumbopelvic kinematics within research settings [15, 16, 220]. They are not applicable in daily clinical practice, however, due to their high cost, required installation space, specific marker placement, and the need for subsequent data capture, analysis, and processing. These factors limit the analysis to standard procedures [17]. Clinicians need alternative measurement systems that are objective, valid, and reliable so they can assess and monitor individual patient changes and compare different population groups. To overcome these limitations, wireless movement-analysis systems using sensors worn on the body have recently been developed (e.g., Valedo® from Hocoma AG, ViMove from dorsaVi, or Reablo® from Corehab). These clinical systems comprise multiple small, lightweight IMUs, which measure the angular tilt, acceleration, and velocity of body segments with respect to magnetic fields and gravity [221]. By combining the output of multiple IMUs and post-processing algorithms into an IMU system, it is possible to estimate lumbopelvic kinematics noninvasively.

To assess the functionality of people with NSLBP, their lumbopelvic kinematics can be quantified [219] (see chapter 4.4). Specifically, assessments can address lumbar range of motion (ROM), movement smoothness, movement control, proprioception, and movement variability and complexity [9, 219, 222-225]. The analysis of lumbar ROM, movement smoothness, movement control and proprioception requires linear indices of lumbopelvic kinematics. ROM is calculated from angular displacement as are indices of movement control and proprioception [219, 225]. Furthermore, derivates of angular displacement such as angular velocity, acceleration, and jerkiness can be used to quantify lumbopelvic kinematics. For instance: root mean squared angular jerkiness is a measure of movement smoothness [222]. Well-known indices such as standard deviation or coefficient of variability are common measures of movement variability [224]. They indicate the size of movement variability but not its structure; thus, they provide only limited information on the use of the redundancy of the neuromuscular system and neuromuscular integrity. Indicators of the latter are the degree of structure in the variability and entropy contained in a movement, quantified through nonlinear indices such as sample entropy and recurrence quantification.

2.4.2 Validity and Reliability

Together with a sound theoretical concept, validity and reliability are prerequisites for a measurement system to have scientific and clinical relevance. Validity is the degree to which it measures what it is supposed to. Reliability is the extent to which a measurement gives consistent results [226]. The concept of validity encompasses

different measurement properties, such as criterion, content, and construct validity [227]. Criterion validity assesses whether a novel measurement system has properties expressed as concurrent and predictive validity. Concurrent validity concerns the degree to which a novel measurement system correlates with a validated measure. Correlation studies between two systems should provide both a measure of random error, or precision, as well as of the accuracy of the devices in their units of measurement (e.g., degrees) [228]. In a systematic review of the literature, Cuestas-Vargas and colleagues found that IMU systems can be concurrent to optoelectronic analysis of trunk measurements, but the degree of concurrent validity is specific to the IMU system and anatomical site [15]. Reliable measures of lumbopelvic kinematics are needed to monitor individual changes over time and to compare different individuals. Concerning the degree to which repeated measures provide similar results, reliability is affected by interrater, intrasession, and intersession variability [228, 229]. The first type is unlikely to be a concern for measurements with an IMU system, except for sensor placement. Variability of sensor placement can be minimised by using a standardised protocol [230]. Intra- and intersession variability depend on biological variability; hence, they are test specific, and the magnitude of their variability indicates the reliability of tests. Furthermore, they can be used to make recommendations for the number of trials needing to be averaged from one or more sessions to improve reliability [231].

Reliable indices of lumbopelvic kinematics measured with a concurrently valid IMU system could be used to investigate unresolved questions regarding the relationship between lumbopelvic kinematics and lumbar spine function. Examples include the effects of NSLBP intensity, fatigue, and exercise therapy on lumbopelvic kinematics.

3 AIMS

This doctoral thesis introduces a novel, wireless movement-analysis system based on IMUs. It focuses on the concurrent validity of the novel IMU system and the reliability of lumbopelvic kinematics measured with it. Furthermore, this thesis explores the effects of NSLBP intensity, fatigue, and exercise therapy on lumbopelvic kinematics.

In using a novel IMU system, the aims of the studies conducted as part of this thesis were to assess the following:

1. The concurrent validity of lumbopelvic kinematics (Study I),
2. The reliability of lumbopelvic kinematics (Study II),
3. The associations between NSLBP intensity and lumbopelvic kinematics (Study III),
4. The associations between fatigue and NSLBP with non-linear, lumbopelvic kinematics (Study IV), and
5. The effect of NME on non-linear, lumbopelvic kinematics (Study V).

4 METHODS

This doctoral thesis included four laboratory studies (Studies I–IV), conducted at the movement laboratory of the Zurich University of Applied Sciences, Winterthur, Switzerland, and one intervention study (Study V) conducted at the UKK Institute, Tampere, Finland.

4.1 Study Population

Thirty-one painfree participants (Studies I–IV) and 21 (Study II), 63 (Studies III and IV), and 83 (Study V) participants with subacute to chronic NSLBP were recruited from local physiotherapy practices, a rehabilitation centre, the campus, and through newspaper advertisements in Winterthur, Switzerland (Studies I–IV) and from the female nursing personnel of Tampere University Hospital in Finland (Study V). Table 1 provides inlusion and exclusion criteria. All studies were conducted according to the declaration of Helsinki and were approved by the local ethics committees of the canton of Zurich, Switzerland, and of Tampere University, Tampere, Finland (KEK-ZH-2011-0522 and R08157). Participants signed written informed consent forms.

Table 1 Inclusion and Exclusion Criteria: ODI = Oswestry diability index; LBP = low back pain; NME = neuromuscular exercise; NRS = numeric rating scale; NSLBP = nonspecific low back pain

	Studies I–IV		**Study V**
	Painfree Participants	**Participants with NSLBP**	**Female Nurses with NSLBP**
Inclusion Criteria	• Age 18–65 years	• Age 18–65 years • NSLBP > 4 weeks • ODI > 8% • StarT Back psychosocial subscale < 4	• Age 30–55 years • Work in current job ≥ 12 months • NSLBP intensity ≥ 2 points on NRS (range 1–10 points)
Exclusion Criteria	• NSLBP • Vertigo or disturbance of the equilibrium • Systemic diseases • Pain in other body areas • Complaints, injury, or surgery of the lower extremities in the past 6 months • Pregnancy	• Specific LBP • Vertigo or disturbance of the equilibrium • Systemic diseases • Pain in other body areas • Complaints, injury or surgery of the lower extremities during the past 6 months • Pregnancy	• Serious former back injury • Chronic LBP defined by a physician • Self-reported continuous NSLBP of ≥ 7 months duration • Diseases or symptoms that limit participation in moderate intensity NME • Current engagement in NME more than once a week • Pregnancy • Postpartum < 12 months

4.2 Measurement Systems

We measured thoracic, lumbo-plevic, and hip kinematics using the novel IMU system (Valedo® Hocoma AG). Additionally, we used an optoelectronic motion capture system (VICON, Oxford UK) during Study I.

4.2.1 Inertial Measurement System

The Valedo® IMUs contain a tri-axillar gyroscope, magnetometer, and accelerometer, as well as a wireless antenna and signal processing unit. Their accuracy is specified at ±0.1° over a range of 360° around all axes [232]. Following a reliable procedure to determine spinal landmarks by their relative distance to C7 and the midpoint between the posterior iliac spines, IMUs were placed on the right thigh (THI), over the sacrum (S2), and at the levels of L1 and T1 [230, 233]. The IMUs were mounted on a plastic frame and attached to the skin with hydrogel tape (KCI Medical GmbH, 8153 Rümlang, CH). The IMU sensor data were transmitted to a recording computer at 200Hz (Studies I, III, and IV) and 50Hz (Studies II and V), respectively. Hocoma AG provided custom data acquisition and synchronisation software (Valedo® Research). The raw IMU sensor data were transformed into quaternions according to Madgwick and colleagues [234] and filtered using a second-order, zero-phase, low-pass Butterworth filter (6Hz cutoff frequency). The angular difference between two IMUs placed above the body segments was calculated using the tilt/twist formulation to prevent rotational singularities [235], with sagittal and frontal planes defined by the global coordinate system. The following sign convention was adopted: flexion, lateral flexion towards the right, and axial rotation towards the left were assigned positive values. An alignment of the IMUs defined the angle of zero degrees. The angle between the L1 and T1 segments was termed "thoracic spine", the angle between S2 and L1 "lumbar spine", and the angle between thigh and S2 "hip". Angular velocity, acceleration, and jerkiness were calculated using the first, second, and third derivative of the filtered angular displacement data. Bauer and colleagues (2015) provide a complete description of the data processing, from raw data to tilt/twist angles [236].

4.2.2 Optoelectronic System

The optoelectronic system consisted of 12 infrared cameras and reflective markers placed above and below every IMU, with a third marker attached to the stiletto on the plastic frame. The coordinate system of each segment, defined by three reflective markers, was aligned to the coordinate system of the IMU. The IMU and optoelectronic systems were synchronized using digital signals generated from a Labjack U3® data-acquisition device (Labjack Corporation, USA). Data were sampled at 200Hz and processed using VICON Nexus® software. The segmental kinematics were calculated as described above. Figure 8 illustrates the IMU and marker placement.

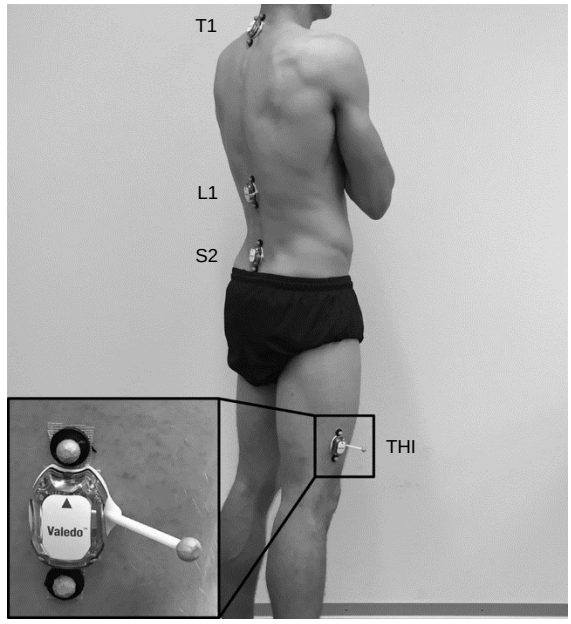

Figure 8. Experimental setup with IMU placement and reflective markers. From top down: T1 = on thoracic vertebra, T1; L1 = on lumbar vertebra, L1; S2 = on sacral vertebra, S2; THI = on the lateral side of the right thigh. Taken with permission from Elsevier.Ltd [236].

4.3 Movement Tests

In summary, 14 movement tests were used in this doctoral thesis. Figure 9 and appendix 10.1 provide a detailed overview of the tests used.

Active ROM tests measure the flexibility of the participant's spine. They combine tests of local joint ranges, movement control, muscle power, and the patient's willingness to perform a movement. Four frequently used ROM tests were included [11].

Active MCI tests evaluate a person's ability to control and differentiate movement between two body segments and to stabilize the spine. A battery of six MCI tests, which form an integral part of one classification system was included [9].

Repeated movement (RM) tests are a rather novel method used to simulate occupational settings that feature heavy, repetitive physical work such as lifting, and they measure a person's variability when performing the same movement repetitively. Two RM tests were included.

Reposition error (RE) tests evaluate a person's proprioceptive deficits within the spine and typically involve participants trying to reproduce a specific target body position. Two RE tests were included.

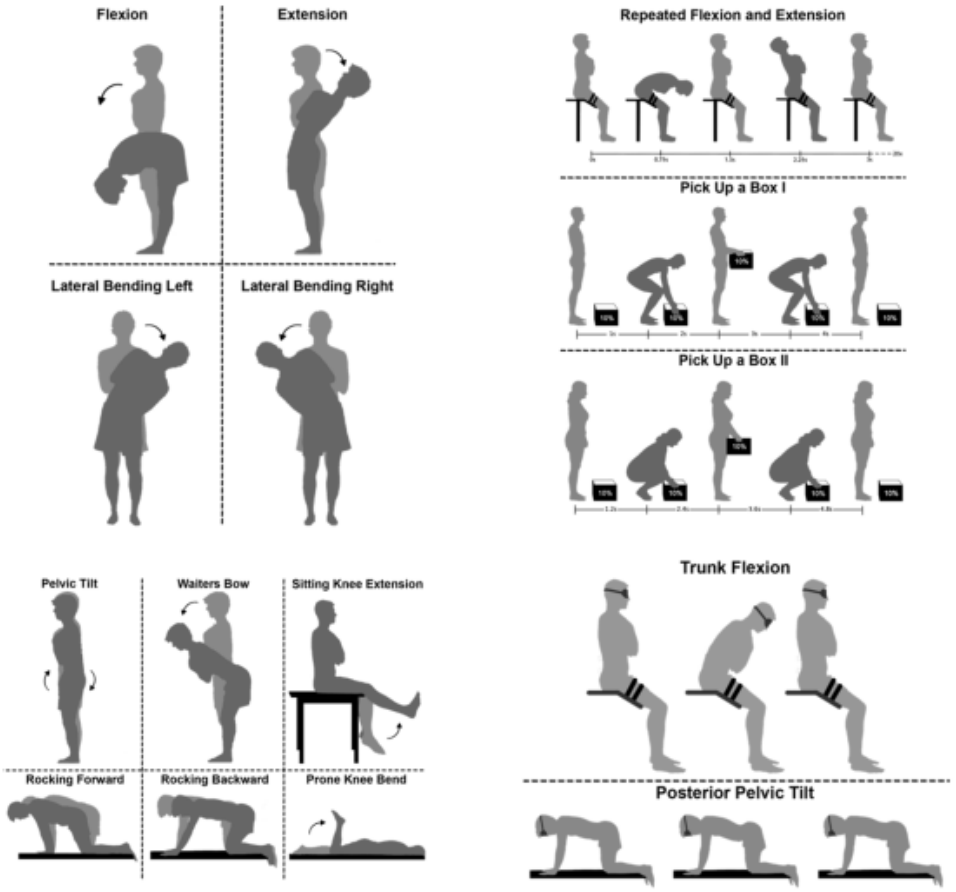

Figure 9. Movement Tests; Top Left: Range of Motion Tests; Bottom Left: Movement Control Impairment Tests; Top Right: Repetaed Movement Tests; Bottom Right: Reposition Error Tests. Pick Up a Box I was used in studies II & III, Pick Up a Box II was used in study V. Extracts of this figure are taken with permission from Elsevier.Ltd [237, 238].

4.4 Lumbopelvic Kinematics

All data processing and calculations were done using Matlab 2012b® (Mathworks, USA), including recurrence quantification code from University of Potsdam, Germany [239] and sample entropy code from Nanyang Technological University, Singapore [240]. Table 10 provides an overview of the included outcome variables per test.

4.4.1 Linear Indices

4.4.1.1 Range of Motion

Linear indices quantify movement during the ROM, MCI, and RE tests. ROM tests were analysed by calculating the ROM of each body segment,

$$ROM = \alpha_{max} - \alpha_{min}$$

with α_{max} and α_{min} being the maximal and minimal angles during the movement.

The ratio of the ROM of the stabilized over the moving body segments and the ROM of the moving segment quantify movement control during MCI tests.

4.4.1.2 Root Mean Squared Jerk

The root mean squared jerk (rmsj) is an indicator of movement smoothness [222, 241]

$$rmsj = \sqrt{\overline{{}^{\circ}/s^3}^2}$$

with $^{\circ}/s^3$ being jerk, or the third derivate of angular displacement.

4.4.1.3 Constant Error

RE tests were evaluated using constant error (*CE*), or the signed difference between the starting (α_{start}) and final angle (α_{fin}), expressed as

$$CE = E[\alpha_{fin} - \alpha_{start}]$$

with E being the expected error which is equivalent to the mean error in finite populations [225].

4.4.2 Nonlinear Indices

Recurrence (REC), determinism (DET), and sample entropy (SaEn) quantify the intrinsic structure of movement variability during RM tests. REC, DET, and SaEn describe different aspects of a time series signal: REC measures the probability of recurrence of movement patterns. Lower REC implies a lower probability that a specific state will recur. DET indicates the predictability of a signal by providing an indication of the structure in variability. Lower DET implies a lower degree of structure in the signal's variability. SaEn is a measure of complexity. As such, the future state of the time series is less predictable. Lower SaEn indicates that a signal is less complex. Movement data are projected into a phase space by taking time-delayed samples from them to calculate REC, DET, and SaEn.

4.4.2.1 Recurrence Quantification Analysis

The time-delayed samples represent movement patterns that can be visualized as points in the phase-space plot. The vectors consisting of the time-delayed samples are called embedding vectors. In recurrence quantification analysis, similar movement patterns are located close to each other, forming a cluster of recurrent points ($R_{i,j}$) [242]. The similarity of movement patterns is quantified by calculating the Euclidean distances between the embedding vectors. In this thesis, the phase-space reconstruction was undertaken separately for angular displacement, velocity, and acceleration data by using a set of input parameters (Appendix 10.2). The delay was estimated using mutual information analysis. The first minimum of mutual information defined the optimal delay. The embedding dimension was estimated by calculating the correlation dimension with different embedding dimensions. The starting point at which the correlation

dimension did not increase significantly, although increasing the embedding dimension, defined the embedding dimension value. The recurrence threshold was defined as 1.3 times the standard deviation of the phase space trajectory. The optimal minimal length of the diagonal lines was chosen after visual inspection of the recurrence plots. All $R_{i,j}$s were subsequently transferred into an $N \times N$–sized recurrence plot (RP), with N being the number of points in the reconstructed state space. *REC* is a measure of the density of the recurrent points in the RP, expressed as

$$REC = \frac{1}{N^2} \sum_{i,j=1}^{N} R_{i,j} * 10^2$$

DET is the amount of $R_{i,j}$s that form diagonal lines (i.e., that are sequential to each other in time) of a prespecified minimal length (*lmin*; Table 11) expressed as

$$DET = \frac{\sum_{l=lmin}^{lmax} l * P(l)}{\sum_{l=1}^{lmax} l * P(l)} * 10^2$$

with *l* being the length of the diagonal lines, *lmax* their maximal possible length, *lmin* their minimal length, and *P(l)* the number of diagonal lines of length *l* [243].

4.4.2.2 Sample Entropy

To calculate SaEn, the similarity of movement patterns was quantified by calculating the distances between the embedding vectors as the maximum difference of their corresponding scalar components. The length of created embedding vectors was m and m+1. SaEn was calculated as

$$SaEn = -\ln\left(\frac{\varphi'^{m+1}(r)}{\varphi'^{m}(r)}\right)$$

with φ being the probability that two embedding vectors are similar in comparison to tolerance r (see Table 11) [244].

4.5 Procedures

Participants attended measurement sessions at the movement laboratory at ZHAW, Switzerland (Studies I–IV) and at the UKK Institute, Finland, (Study V) to perform a series of movement tests. Prior to each test, they received standardized oral instructions from one of the examiners in all studies and visual instructions via videos for Studies I–IV. In case of poor initial performance, the examiners repeated the instructions up to three times and demonstrated the test. If one participant was still performing a test incorrectly, it was permitted.

4.5.1 Concurrent Validity of Lumbopelvic Kinematics (Study I)

Participants attended one measurement session and performed the four ROM tests three times in randomized order. These tests provide information on the concurrent validity of a system over a large ROM, and allow for differentiation between the primary movement direction (main movement) and secondary ones (adjunct movements). The ROMs for the thoracic spine, lumbar spine, and hip were calculated.

4.5.2 Reliability of Lumbopelvic Kinematics (Study II)

Participants attended two identical measurement sessions, separated by a one-week period. Both sessions took place at the same time of day. Participants performed all 14 tests seven times, except for those in four-point kneeling (4pk), which was repeated five times to minimise stress to their wrists. To reduce stress on their backs, the participants with NSLBP did all tests five times and omitted the RM tests. The order of the tests was randomized between participants but not between days. The participants were given a rest period of 30 seconds maximum between the tests, but they were allowed to continue earlier at their own convenience. For the RM, MCI, and RE tests, the linear indices described above were calculated. Table 11 describes the input parameters used to calculate REC, DET, and SaEn from the RM tests.

4.5.3 Associations Between NSLBP Intensity and Lumbopelvic Kinematics (Study III)

Participants attended one measurement session in which they performed the MCI tests "Sitting Knee Extension" and "Waiter's Bow" three times and the RM test "Pick Up a

Box I" once. The latter consisted of ten cycles lasting four seconds, starting in upright standing. The participants chose the rest time between the tests.

For "Sitting Knee Extension", the ROM at the lumbar spine (ROMLS) was calculated. For "Waiters Bow," the ratio between the ROM of the lumbar spine and the hip ($\frac{LS}{Hip}ROM$) was calculated. For all outcomes, the mean of the three repetitions was used for statistical analysis. Table 11 describes the input parameters used to calculate REC and DET.

4.5.4 Associations Between Fatigue and NSLBP with Lumbopelvic Kinematics (Study IV)

Participants attended one measurement session and performed the RM test "Repeated Flexion and Extension" before and after fatiguing the lumbar musculature. It consisted of 20 cycles three seconds long, starting in upright sitting. The duration of the complete test was 60 seconds, and it was performed once, pre- and post-fatiguing. Participants performed the posttest immediately after completing an isometric endurance test to fatigue the lumbar musculature. Table 11 describes the input parameters used to calculate DET and SaEn.

As it is described elsewhere [245], the isometric endurance test is summarized briefly here. Lying in prone position on a physiotherapy bench tilted 45° downwards from the pelvis, participants were instructed to lift their upper bodies from the bench and maintain them unsupported for as long as possible. Their lower legs and thighs were stabilized with two belts. The test ended when they could no longer maintain this position or if an investigator noticed their upper body touching the bench, with the posttest beginning immediately after. All participants received standardized verbal encouragement during the endurance test.

4.5.5 Effect of Neuromuscular Exercise on Lumbopelvic Kinematics (Study V)

Study V was a planned secondary analysis of the third substudy of the NURSE-RCT [246]. The original study was a four-arm, randomized clinical trial set up as follows: (1) combined NME and back counselling (2) NME only, (3) back counselling only, and (4) a nontreatment control.

4.5.5.1 Randomization

Once participants had consented to enter the study, sequentially numbered sealed envelopes were used to assign them to four study groups. At baseline measurements, an envelope next in order was opened and he or she was then offered the allocated study group, as well as information relevant to practical participation. The personnel conducting study measurements were blind to group allocation, as were the statistician and outcome assessors (until completion of statistical analysis). In Study V, the participants in arms 1 and 2 were the NME intervention group and those in arms 3 and 4 the control group.

4.5.5.2 Intervention

Organised near the workplaces of the nurses, the NME intervention lasted for six months. Participants in this group were expected to train twice a week in 60-minute sessions. The overall aim was to reduce pain-induced disturbances of movement control, while increasing strength and endurance. During the first seven weeks, participants practiced correct performance techniques, control of their lumbar neutral zone, and specific breathing techniques. The program was progressive as the demands on coordination, balance, strength, and endurance increased. The training principles and key exercises (which included "Bird Dog" and "Shoulder Bridge") are described in detail in the study protocol (online Supplementary appendix 1) by Suni and collegues [246]. Educated, experienced leaders taught the exercise sessions. Participants were expected to participate in instruction sessions twice a week during the first two months, and in the following four months they had one instructed session and one home session per week (with the help of a DVD and booklet produced for the study that included the training principles and key exercises). After the six-month intervention, subjects were encouraged to continue exercising at home. Furthermore, two instructed sessions were provided at the beginning of the active follow-up period. The control group only attended the three measurement sessions (baseline, 6 months, and 12 months).

4.5.5.3 Movement Test

Starting in upright standing, participants performed the RM "Pick Up a Box II" three times. It consisted of five cycles, each lasting 4.8 seconds (Figure 9). DET was calculated

as described above (Appendix, Table 11). The mean of the three repetitions was used for further analysis.

4.5.6 Grouping Variables and Covariates

NSLBP intensity was used as a grouping variable in Studies II–IV and as a covariate in Study V. In Studies II–IV, participants rated their NSLBP intensity, defined as the mean level of NSLBP during the past four weeks, using an 11-point numerical rating scale (NRS) anchored with "no pain" (0) and proceeding to "the worst pain imaginable". In Study III, participants were allocated to one of eight groups according to their perceived NSLBP intensity. In Study IV, they were assigned to a painfree group (NRS = 0) and an NSLBP group (NRS = 1–10). In Study V, NSLBP intensity at baseline — defined as the mean level of NSLBP pain during the past four weeks — was measured using a visual analogue scale (VAS) anchored with "no pain" (0 mm) and "the worst possible pain imaginable" (100 mm). Age, gender, and body mass index (BMI), which can influence lumbopelvic kinematics, were recorded as covariates in Studies III–V [247]. Physical stress at work (PS) was measured with a five-point Likert scale ranging from "almost no physical stress" (1) to "maximal physical stress" (5) and was recorded as a covariate for Study III [248].

4.6 Statistical Analysis

All statistical analyses were conducted using R (R Foundation for Statistical Computing, Austria).

4.6.1 Concurrent Validity of Lumbopelvic Kinematics (Study I)

The coefficient of determination (r^2), a measure of precision, and the root mean squared error (RMSE), a measure of accuracy, quantify the concurrent validity of the IMU system. The values of r^2 range from 0 to 1. A high value of r^2 implies that angles measured by IMUs and the optoelectronic system have the same characteristic. RMSE is the measure of the average difference between the two signals, in the original scale. Systematic differences between the systems were analysed using the Wilcoxon rank-sum test, with p set at <0.05.

4.6.2 Reliability of Lumbopelvic Kinematics (Study II)

The generalizability theory [249] with the design $p \times t \times d$ ($participants \times trials \times days$) was used as a framework to estimate the reliability of lumbopelvic kinematics. The index of dependability Φ was calculated to assess the diagnostic value of an index, and reliability was interpreted as <0.25 = very low; 0.26–0.49 = low; 0.50–0.69 = moderate; 0.70–0.89 = high, and >0.90 = very high [250]. Φ≥0.70 was interpreted as sufficient to compare between different individuals. Subsequently, Φ coefficients were calculated for alternative measurement strategies, where the number of trials was varied up to ten, and the number of days up two, which represent acceptable measurement strategies. The number of required trials per day necessary to achieve high reliability was evaluated thereby. Calculated to assess the ability of an index to detect changes over time, the coefficient of variation (CV) was rated as follows: >10% not reliable, 6–10% adequately reliable and 5% highly reliable [251]. CVs ≤ 10% were construed as sufficient to monitor changes over time [252].

4.6.3 Associations Between NSLBP Intensity, Fatigue, and Effect of Neuromuscular Exercise on Lumbopelvic Kinematics (Studies III-V)

Linear mixed models were fitted for each index, with the quantities of interest being NSLBP, fatigue NSLBP, or treatment effect. An NSLBP intensity effect in Study III would indicate that NSLBP intensity affects lumbo-plevic kinematics. A fatigue NSLBP effect in Study IV would indicate that one group responds differently to fatigue compared to the other. A treatment effect in Study V would indicate that NME influenced lumbopelvic kinematics compared to no intervention.

In Study III a stepwise model selection procedure with backwards optimisation by the Akaike-Information-criterion was used to determine the effect of NSLBP intensity on lumbopelvic kinematics. We created point estimates of NSLBP intensity and 95% confidence intervals (95%CI). A 95%CI not crossing zero indicates a significant NSLBP intensity effect.

In Studies IV-V, a Bayesian estimation of the model parameters was performed by using uninformative priors on the model parameters. We used Monte-Carlo-Markov-Chain (MCMC) algorithms with Gibbs sampling to sample from the posterior distributions [253] Fatigue (Study IV), and treatment effects (Study V), as well as 95% highest posterior density intervals (95%HPDI) were estimated. The probability that the

magnitude of an effect is within the borders of the 95%HPDI is 0.95. The 95%HPDI is thus comparable to the 95%CI in frequentist statistics. In Study V, the analysis followed the "modified intention to treat" principle: all participants were analysed based on the initial treatment assignment. Missing data from dropouts were not imputed. We assumed that the missing mechanism was at random (MAR), according to Little and Rubin [254]. We did not use simple solutions such as "Carrying Forward", as these are not recommended [255]. The statistical analysis can be explained in more detail using the example of Study V: For each outcome, a linear mixed model was fitted to the data. The model for observation Y_{ijk}(k_{th} participant in the i_{th} group at time j) was

$$Y_{ijk} = \mu + \alpha_i + \beta_j + (\alpha\beta)_{ij} + U_{k(i)} + \beta_{covariates} + \varepsilon_{ijk},$$

with α_i as the i_{th} group effect, β_j as the j_{th} time effect, $(\alpha\beta)_{ij}$as the ij_{th} group-time interaction (corresponding to the treatment effect, the quantity of interest), $U_{k(i)}$ as the random intercept of subject k nested in group I, $\beta_{covariates}$ as the effect of BMI, age and LBP intensity at baseline, and ε_{ijk} as measurement error. We assumed $U_{ik} \sim N(0,v^2)$, with v^2 as the between-subject variance and ε_{ijk} $N(0,\tau^2)$ with τ^2 as the within-subject variance. A Bayesian estimation of the model parameters was performed by using uninformative priors on the model parameters. We used Gibbs sampling [253], a Monte-Carlo-Markov-Chain (MCMC) algorithm, to sample from the posterior distributions. Means, standard deviations and 95% highest posterior density intervals (95% HPDI) were reported from the simulated posterior distributions. $(\alpha\beta)_{ij}$ were calculated as

$$(\alpha\beta)_{ij} = \delta_B - \delta_A$$

with δ_A respectively δ_B calculated as

$$\delta_A = Group\ A_{post} - Group\ A_{pre}$$

with $Group\ A_{post}$ being the observed mean value of Group A after an intervention and $Group\ A_{pre}$the observed mean value of Group A before an intervention (e.g., intervention group A had a mean value of 65.6 before and 63.4 after an intervention, while control group B had a mean value of 63.7 before and 67.3 after). Therefore,

$$\delta_A = 63.4 - 65.6 = -2.2$$
$$\delta_B = 67.3 - 63.7 = 3.6$$
$$(\alpha\beta)_{ij} = 3.6 - -2.2 = 5.8$$

5 RESULTS

The ratio of females and males recruited, as well as their mean age, BMI, and NSLBP intensity, were equivalent through all of the studies except Study V, in which the participants were female only, on average older, and had a greater BMI. The control group also reported lower NSLBP intensity at baseline. Table 2 illustrates the participants' characteristics. This thesis book includes the main results of studies I-V. The additional data are presented in the articles at the end of the book [236-238, 256, 257].

5.1 Concurrent Validity of Lumbopelvic Kinematics (Study I)

The measurement systems showed acceptable agreement and small measurement errors in the primary movement direction (Table 3). The r^2 coefficients ranged from 0.94–0.99, except for hip movement during the lateral flexion tests (0.85–0.87). Flexion of the lumbar spine and hip, as well as lateral flexion of the thoracic and lumbar spine, revealed very high agreement, with an r^2 coefficient of 0.99. The RMSE ranged between 1.1 and 6.1°. In the nonprimary movement directions, r^2 coefficients were lower (0.36–0.87). The RMSE were similar (1.2–6.8°) compared to the primary movement direction and were therefore greater relative to the measured ROM (Table 4). Figure 10 illustrates representative sample data from the ROM measures.

Based on the results, the analysis during Studies II–V focused on the primary movement direction of each test.

5.2 Reliability of Lumbopelvic Kinematics (Study II)

Tables 5 and 6 summarize the grand mean, Φ-coefficients, and number of trials averaged from one or two measurement days needed to gain $\Phi \geq 0.70$ and the CV for the MCI and RM variables, since those tests were used in Studies III–V. The reliability data for all variables are provided in the articles at the end of the book [236, 257]. In general, ROM and RM tests required a smaller number of trials to reach high reliability

and had smaller CVs compared to MCI and RE tests. ROM tests revealed high to very high reliability using a single trial only on one day — except for extension of the lumbar spine and, in participants with NSLBP, both lateral bending tests. All CVs were smaller than 10%. The RM tests showed CVs smaller than 10%, with "Picking up a Box" being more reliable than the "Flexion and Extension" test. The MCI tests differed in their reliability, with Φ-coefficients of a single measurement ranging from low to high, and CVs from 8 to 84%. The RE tests showed low reliability for a single measurement, with CVs greater than 10%.

Based on these results, it was decided to use the RM "Pick Up a Box" test with two repetitions of five cycles in Study III. (They were collapsed into one repetition of ten cycles to minimize stress on the participants with NSLBP.) To improve reliability further, 15 cycles were used in Study V. The RM "Repeated Flexion and Extension" test was performed with four repetitions of five cycles in Study IV. (These were collapsed into one repetition of 20 cycles to minimize stress on the participants with NSLBP.) The MCI tests "Waiter's Bow" and "Sitting Knee Extension" were performed with three repetitions in Study III. "Waiter's Bow" was selected because $\frac{LS}{Hip}ROM$ showed acceptable reliability in both populations. "Sitting Knee Extension" was selected because $ROM\ LS$ showed acceptable reliability for pain-free participants and because tests to investigate both flexion movement control impairment ("Waiter's Bow") and extension movement control impairment ("Sitting Knee Extension") were of interest [258].

5.3 Associations between NSLBP Intensity and Lumbopelvic Kinematics (Study III)

NSLBP intensity affects lumbopelvic kinematics. Participants with higher intensity NSLBP showed less recurrent and less deterministic movement during the RM test. This means that their movements were more variable and less predictable. The observed effects for a one-point increase in NSLBP were: REC decreased –0.25 and –2.3 for angular velocity and acceleration, respectively. For angular displacement, velocity and acceleration, DET decreased –0.06, –2.3 and –0.86, respectively.

NSLBP intensity had no effect on $ROM\ LS$ or $\frac{LS}{Hip}ROM$ during the MCI test. **Table 7** illustrates the descriptive statistics and NSLBP effects with corresponding 95%CI. Table 12 in the Appendix illustrates the final models for each outcome. Appendix 10.2 depicts angular displacement and the corresponding recurrence plot measured during

the "Pick Up a Box" test of one representative painfree participant and one participant with high-intensity NSLBP.

5.4 Associations between Fatigue and NSLBP with Lumbopelvic Kinematics (Study IV)

Fatigue affects lumbopelvic kinematics. This effect depends on the presence of NSLBP. The painfree participants showed more complex and less predictable lumbar movement after an isometric endurance test than participants suffering from NSLBP.

This fatigue NSLBP effect was observed in angular velocity, but not in angular displacement. Additionally, DET and SaEn of angular velocity decreased (–0.2) and increased (0.1) per year of a participant's life, indicating a minor effect of age. Gender and BMI had no effect. **Table 8** illustrates the descriptive statistics and fatigue NSLBP effects, with corresponding 95%HPDI. Figure 11 illustrates the pre post differences between the groups.

5.5 Effect of Neuromuscular Exercise on Lumbopelvic Kinematics (Study V)

NME affects lumbopelvic kinematics. When compared to no NME, it may reverse or reduce deterioration of lumbar movement, by decreasing or preserving the degree of structure of the movements' variability. This means that lumbar movement was less predictable following the NME intervention; or, respectively, that the degree of predictability was preserved.

In the NME group, DET of angular displacement decreased and DET of angular velocity remained constant during the intervention phase. Both increased in the control group. The observed treatment effect is substantial; as its magnitude is approximately one-half of the standard deviation of both outcomes. **Table 9** illustrates the descriptive statistics and treatment effects, with corresponding 95%HPDI. Figure 12 illustrates the pre post differences between the groups.

Table 2. Participant characteristics for Studies I –V. Results are provided as mean (± standard deviation) or median (range); BMI = body mass index; f = female; LBP = low back pain; m = male; n = number of participants; NME = neuromuscular exercise; NSLBP = nonspecific low back pain; NRS = numeric pain rating scale; ODI = Oswestry Disability Index; PS = physical stress at work; VAS = visual analogue scale.

Studies	Group	N	Gender (f/m)	Age (years)	BMI (kg/m^2)	NSLBP Intensity (NRS 0–10 or VAS mm)	ODI %	PS
Study I	Painfree	22	11/11	41.2 (±11.1)	22.9 (±2.9)			
Study II	Painfree	24	13/11	38.0 (± 11.2)	22.9 (±2.7)			
	NSLBP	21	4/17	33.9 (±12.1)	24.9 (±6.4)	2.8 (±1.3) NRS	15.9 (±4.1)	
Study III	Painfree	31	17/14	40.1 (±12.1)	22.7 (±2.9)			1 (1–4)
	NSLBP	63	31/32	39.2 (±12.6)	24.2 (±4.0)	3.3 (±1.5) NRS	17.5 (±7.0)	1 (1–5)
Study IV	Painfree	27	15/12	39.6 (±11.6)	22.7 (±2.8)			
	NSLBP	59	29/30	39.1 (±12.8)	24.0 (±3.6)	3.4 (±1.7) NRS	18.1 (±7.0)	
Study V	NME	42	42/–	45.7 (±7.8)	26.7 (±4.6)	34.0 (±21.0) VAS *		
	Control	41	41/–	46.7 (±7.7)	25.8 (±3.6)	28.0 (±21.1) VAS *		

* at baseline.

Table 3. Study I: Concurrent validity of the IMU system in the primary movement direction. *indicates a significant systematic difference between the two systems. Results are provided as mean (±SD). IMU = inertial measurement unit system, Opt = Optoelectronic System, r^2 = R-squared; RMSE = root mean squared error; ROM = range of motion; SD = standard deviation. Taken with permission from Elsevier.Ltd [236].

Test	**ROM Thoracic Spine, °**				**ROM Lumbar Spine, °**				**ROM Hip, °**			
	IMU	Opt	r^2	RMSE	IMU	Opt	r^2	RMSE	IMU	Opt	r^2	RMSE
ROM Flexion	36.2(±11.9)	29.7(±10.9)	0.95(±0.04)	5.8(±2.0)	53.3(±10.9)	50.71(±9.5)	0.99(±0.01)	4.1(±1.8)	77.4(±15.3)	77.1(±14.2)	0.99(±0.01)	6.1(±2.7)
ROM Extension	22.2(±9.9)	18.9(±9.9)	0.94(±0.09)	5.9(±3.3)	16.6(±10.5)	15.3(±8.4)	0.97(±0.05)	4.4(±2.2)	13.7(±5.8)	14.8(±5.8)	0.94(±0.09)	5.6(±4.1)
ROM Lateral Bending Right	31.9(±5.1)*	35.0(±6.1)*	0.99(±0.01)	2.8(±1.4)	22.8(±5.1)	23.7(±5.1)	0.99(±0.01)	1.8(±0.0)	7.3(±4.3)	7.3(±4.9)	0.87(±0.21)	1.1(±0.7)
ROM Lateral Bending Left	32.6(±9.2)	34.7(±10.3)	0.99(±0.03)	2.6(±2.0)	22.2(±5.7)	22.9(±6.5)	0.99(±0.01)	1.9(±1.3)	6.8(±3.0)	6.9(±3.4)	0.85(±0.20)	1.1(±0.7)

Table 4. Study I: Concurrent validity of the IMU system in the secondary movement directions. * indicates a significant systematic difference between the two systems; Results are provided as mean (± SD); FE = ROM flexion–extension; IMU = inertial measurement unit system, LF = ROM lateral flexion; Opt = Optoelectronic System, RO = ROM rotation; ROM = range of motion; r^2 = R squared; RMSE = root mean squared error; SD = standard deviation. Taken with permission from Elsevier.Ltd [236].

Test	**ROM Thoracic Spine, °**				**ROM Lumbar Spine, °**				**ROM Hip, °**			
	IMU	Opt	r^2	RMSE	IMU	Opt	r^2	RMSE	IMU	Opt	r^2	RMSE
ROM flexion												
LF	15.9(±8.1)	12.7(±6.5)	0.47(±0.31)	4.6(±2.5)	10.4(±4.4)*	8.2(±3.8)*	0.66(±0.27)	5.3(±3.4)	18.1(±7.2)*	23.3(±7.8)*	0.67(±0.37)	6.7(±2.7)
RO	20.6(±8.0)*	14.8(±5.5)*	0.42(±0.34)	6.7(±2.5)	12.8(±6.1)	12.2(±6.7)	0.69(±0.31)	2.6(±1.3)	20.7±(8.1)*	16.7(±5.9)*	0.79(±0.22)	3.9(±3.9)
ROM extension												
LF	5.8(±2.6)	6.3(±2.3)	0.55(±0.27)	2.5(±2.4)	4.9(±2.3)	5.3(±2.4)	0.54(±0.28)	2.1(±1.6)	4.4(±2.5)	4.0(±1.9)	0.45(±0.34)	1.9(±1.0)
RO	8.7(±4.4)	8.3(±5.2)	0.49(±0.31)	3.6(±2.2)	7.6(±3.8)	7.8(±4.5)	0.63(±0.31)	1.9(±1.1)	8.1(±3.1)	8.3(±3.7)	0.77(±0.21)	1.8(±0.8)
ROM lateral bending right												
FE	8.8(± 3.2)	8.6(±3.4)	0.68(±0.24)	5.6(±3.4)	10.7(±3.4)*	9.2(±3.4)*	0.87(±0.17)	4.6(±2.1)	5.0(±2.7)	5.8(±3.3)	0.72(±0.29)	5.3(±3.2)
RO	13.7(± 4.8)	14.1(±5.6)	0.77(±0.24)	2.6(±1.2)	11.2(±4.5)	12.7(±4.6)	0.82(±0.17)	2.1(±1.2)	7.7(±4.5)	6.7(±4.1)	0.81(±0.20)	1.2(±0.7)
ROM lateral bending left												
FE	11.7(±6.4)*	9.7(±5.5)*	0.58(±0.32)	6.8(±4.4)	12.4(±5.2)*	8.9(±4.0)*	0.84(±0.24)	4.9(±2.6)	4.7(±2.3)	4.4(±1.6)	0.75(±0.25)	4.6(±3.1)
RO	16.5(±7.3)	16.8(±5.6)	0.69(±0.29)	5.1(±3.6)	12.9(±4.7)	12.4(±4.8)	0.87(±0.17)	1.7(±0.9)	7.5(±3.2) *	6.3(±2.9)*	0.80(±0.21)	1.3(±0.7)

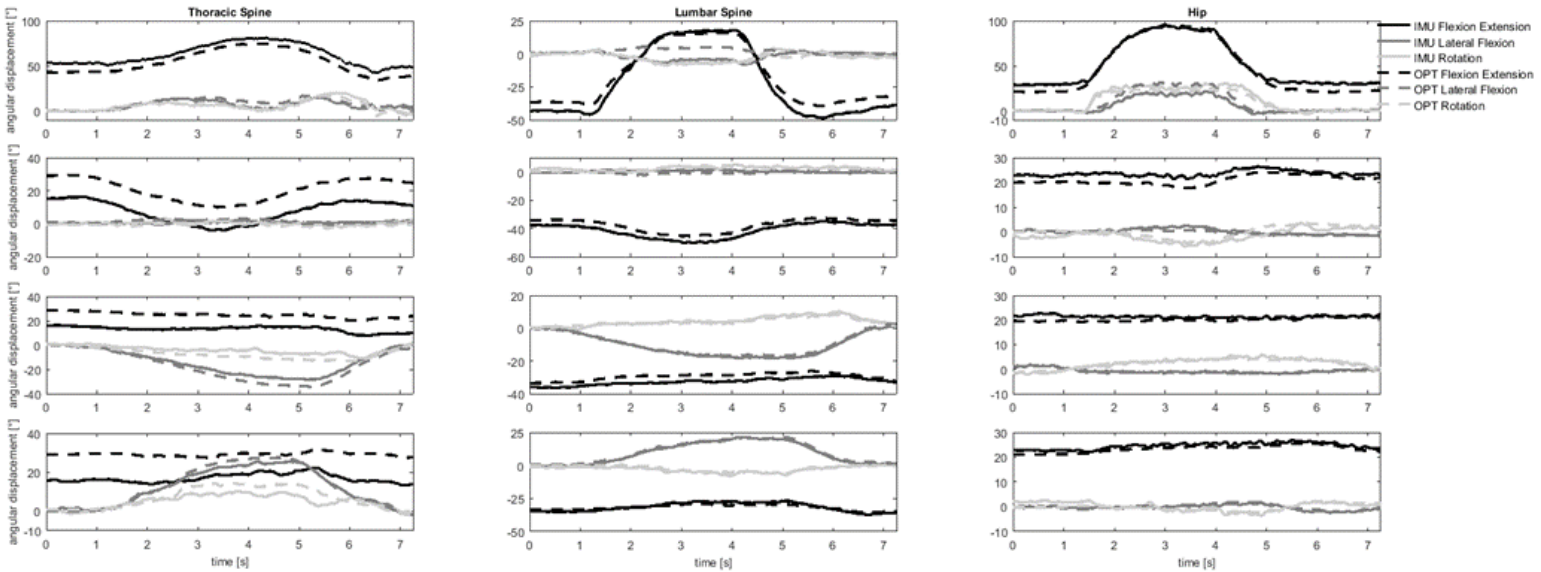

Figure 10. Study I: ROM Tests of Angular Displacement, Thoracic/Lumbar Spine, and Hip. IMU = inertial measurement unit system; OPT = optoelectronic system.

Table 5. Study II: Reliability of indices of lumbopelvic kinematics in painfree participants (reliability of a single measure, number of trials averaged on 1 or 2 days needed to achieve high reliability and coefficient of variation). Φ = index of dependability; DET = determinism; AA = angular acceleration; AD = angular displacement; AV – angular velocity; CV = Coefficient of variation; LS = lumbar spine; MCI = Movement control impairment; PKB = prone knee bend; PT = Pelvic Tilt; PU = Picking Up a Box; RB = rocking backwards; REC = recurrence; RF = rocking forwards; RM = repetitive movement; ROM = range of motion; SaEn= sample entropy; SKE = sitting knee extension; SD: Standard deviation; TC = Thoracic Spine; WB = waiter's bow. Taken with permission from Elsevier.Ltd [236]

Test	Variable; Unit	Mean (±SD)	Φ one trial	No. trials Φ >0.7 1 day	No. trials Φ >0.7 2 days	CV in %
MCI Test						
Pelvic Tilt	Ratio TS/LS ROM	0.2 (±0.1)	0.27	>10	7	16
Sitting Knee Extension	ROM LS (°)	1.9 (±2.8)	0.68	2	1	22
Waiter's Bow	Ratio LS/Hip ROM	0.5 (±0.44)	0.77	1	1	10
Rocking Backwards	Ratio LS/Hip ROM	0.71 (±0.43)	0.38	>10	>10	18
Rocking Forward	Ratio LS/Hip ROM	1.52 (±1.16)	0.19	>10	>10	11
Prone Knee Bend	ROM LS (°)	–4.0 (±2.7)	0.44	>10	3	14
RM Tests						
Picking Up a Box	REC AD	15.4 (±1.3)	0.68	3	2	2
	DET AD	96.7 (±0.7)	0.51	3	2	<1
	SaEn AD	18.9 (±2.9)	0.33	6	3	3
	REC AV	13.7 (±1.4)	0.58	3	2	2
	DET AV	93.6 (±0.9)	0.61	3	2	1
	SaEn AV	16.7 (±2.3)	0.29	8	5	4
	REC AA	12.9 (±1.4)	0.63	4	2	3
	DET AA	74.0 (±4.2)	0.65	3	2	2
	SaEn AA	25.0 (±2.8)	0.38	7	3	5
Repeated Flexion and Extension	REC AD	13.1 (±1.5)	0.29	>10	>10	4
	DET AD	96.6 (±1.3)	0.64	4	3	<1
	SaEn AD	12.9 (±2.0)	0.62	4	3	1
	REC AV	13.5 (±1.8)	0.56	5	4	1
	DET AV	94.5 (±1.1)	0.69	2	1	<1
	SaEn AV	13.1 (±1.7)	0.63	3	2	1
	REC AA	8.8 (±1.1)	0.24	>10	>10	4
	DET AA	66.2 (±7.1)	0.60	6	3	6
	SaEn AA	23.9 (±4.4)	0.28	8	4	5

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Table 6. Study II: Reliability of indices of lumbopelvic kinematics in participants with NSLBP (reliability of a single measure, number of trials averaged on 1 or 2 days needed to achieve high reliability and coefficient of variation). Φ = index of dependability; CV = coefficient of variation; LS = lumbar spine; MCI = movement control impairment; PKB = prone knee bend; PT = pelvic tilt; RB = rocking backwards; ROM = range of motion; RF = rocking forwards; SKE = sitting knee extension; SD = standard deviation; TS = thoracic spine; WB = Waiter's Bow. Taken with permission from Elsevier.Ltd [257].

Test	Variable; Unit	Mean (± SD)	Φ one trial	Number trials Φ >0.7 One day	Number trials Φ>0.7 two days	CV (%)
MCI Tests						
Pelvic Tilt	Ratio TS/LS ROM	0.8 (± 0.4)	0.22	>10	>10	18
Sitting Knee Extension	ROM LS (°)	0.9 (± 1.3)	0.13	>10	>10	67
Waiter's Bow	Ratio LS/Hip ROM	0.7 (± 0.5)	0.91	1	1	8
Rocking Backwards	Ratio LS/Hip ROM	1.2 (± 1.0)	0.30	9	6	37
Rocking Forward	Ratio LS/Hip ROM	2.0 (± 1.4)	0.36	8	4	16
Prone Knee Bend	ROM LS (°)	–1.2 (± 2.3)	0.11	>10	>10	84

Table 7. Study III: Descriptive statistics and NSLBP effects. Results are provided as mean (± SD). Bold numbers indicate the 95% CI did not cross 0. 95CI = 95 % confidence interval; AA = angular acceleration; AD = angular displacement; AV = angular velocity; DET = determinism; LS = lumbar spine; LS/hip ROM = ratio of lumbar spine/hip ROM; NSLBP = nonspecific low back pain intensity; REC = recurrence rate; ROM = range of motion. Adapted and taken with permission from Elsevier.Ltd [238].

NSLBP (NRS 0–10)	**N**	**Sitting Knee Extension** ROM° LS	**Waiter's Bow** LS/hip ROM	**Pick Up a Box** REC AD	DET AD	REC AV	DET AV	REC AA	DET AA
0	31	2.6(±3.7)	0.3(±0.2)	41.0(±1.3)	98.7(±0.4)	43.5(±2.0)	94.0(±3.5)	39.2(±5.1)	58.3(±9.3)
1	4	1.2(±3.8)	0.5(±0.4)	40.1(±0.4)	98.4(±0.4)	42.2(±3.4)	87.1(±9.6)	39.8(±5.9)	55.4(±11.8)
2	19	2.2(±3.0)	0.2(±0.1)	41.7(±2.3)	98.1(±0.5)	43.7(±2.5)	89.6(±6.0)	37.4(±1.9)	52.4(±6.4)
3	13	2.5(±2.7)	0.5(±0.5)	41.4(±1.3)	98.3(±0.4)	42.9(±1.9)	93.2(±3.7)	37.9(±2.6)	54.4(±6.3)
4	15	3.3(±4.0)	0.3(±0.2)	41.9(±1.3)	98.4(±0.3)	42.5(±1.9)	90.6(±3.7)	38.5(±2.6)	52.8(±7.6)
5	5	1.1(±2.7)	0.8(±0.7)	40.9(±0.6)	98.3(±0.2)	41.5(±1.9)	89.4(±4.0)	38.7(±2.8)	53.8(±5.5)
6	4	2.4(±2.7)	0.7(±0.7)	41.4(±1.4)	98.2(±0.3)	42.6(±1.7)	90.4(±3.4)	38.7(±4.4)	51.2(±9.3)
7	3	1.3(±3.5)	0.4(±0.2)	41.9(±1.6)	98.1(±0.3)	41.6(±1.3)	86.7(±6.9)	35.7(±1.4)	49.9(±4.8)
NSLBP Effect		0.3	0.0	0.11	**–0.06**	**–0.25**	**–3.31**	**–2.3**	**–0.86**
95%CI		–0.2 to 0.9	–0.1 to 0.2	–0.5 to 0.26	**–0.11 to –0.02**	**–0.46 to –0.03**	**–6.21 to –0.01**	**–3.87 to –0.2**	**–17.3 to –0.00**

Table 8. Study IV: Descriptive statistics and group fatigue effects. Results are provided as mean (±SD); Bold numbers indicate the 95% HDPI not crossing 0. SAEN values are expressed as $*10^2$. 95%HPDI = 95% highest posterior density interval; AD = angular displacement; AV = angular velocity; DET = determinism; NSLBP = nonspecific low back pain; ROM = range of motion lumbar spine; SaEn = sample entropy. Taken with permission from Elsevier.Ltd [237].

Group	**Fatigue**	**ROM** (°)	**DET AD**	**DET AV**	**SaEn AD**	**SaEn AV**
Pain free	Pre	60.0 (±14.9)	62.8 (±2.9)	36.2 (±8.8)	2.4 (±0.3)	9.7 (±2.4)
	Post	61.7(±21.8)	63.3 (±2.9)	32.1 (±9.3)	2.5 (±0.5)	11.2 (±3.6)
NSLBP	Pre	59.5 (±19.1)	64.6 (±3.9)	34.9 (±8.1)	2.4 (±0.4)	11.7 (±3.4)
	Post	61.6 (±22.4)	64.4 (±4.2)	34.4 (±9.8)	2.5 (±0.6)	11.8 (±3.7)
Group Fatigue Effects			**DET AD**	**DET AV**	**SaEn AD**	**SaEn AV**
Group fatigue effect			–0.7	**3.6**	0	**–1.4**
95%HPDI			–2.4 to 1.0	**2.3 to 7.1**	–0.4 to 0.2	**–2.7 to –0.1**

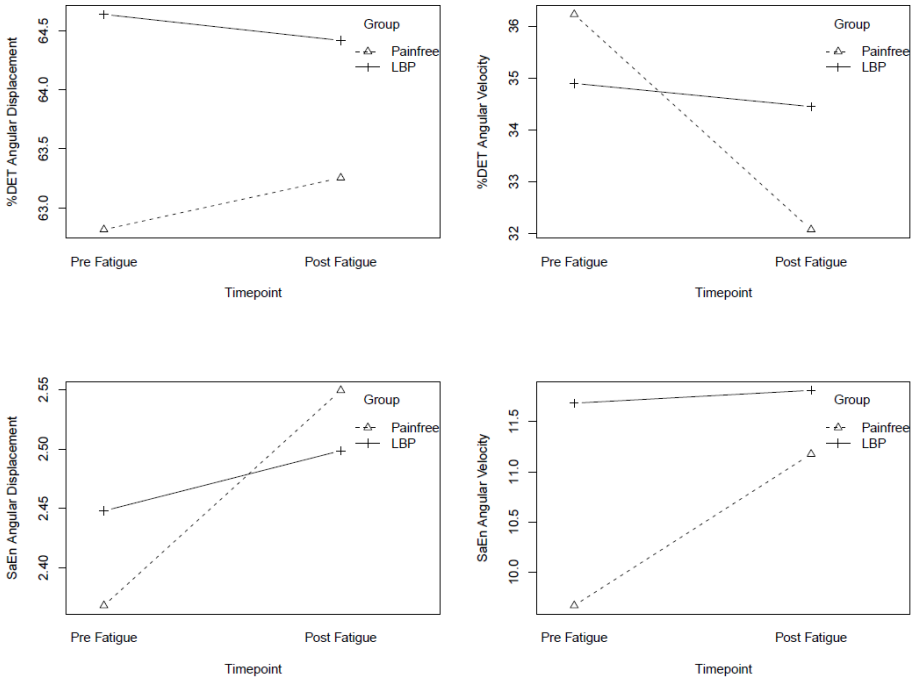
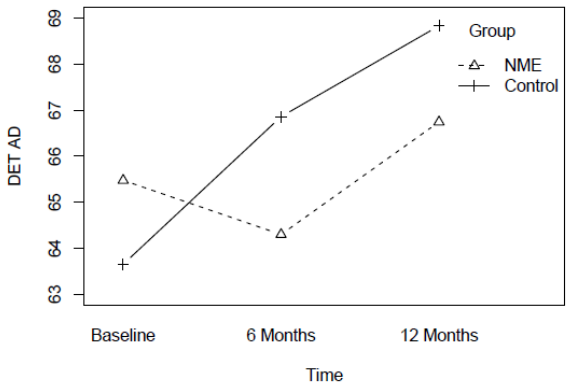


Figure 11 Study IV Interaction plots for the pre post fatigue differences between the groups. DET –determinism; LBP – non specific low back pain; SaEn – Sample Entropy. Note: Figure 11 illustrates the empirical means, controlled for the covariates illustrated in Table 8. These can differ slightly from the observed effects.

Table 9. Study V: Descriptive statistics and treatment effects. Results are provided as mean (±SD). Bold numbers indicate the 95% HDPI not crossing 0. 95%. HPDI = 95% highest posterior density interval; AD = angular displacement; AV = angular velocity; DET = determinism; N = number of participants. Taken from [256].

Group	Time	N	DET AD	DET AV
NME	Pre	42	65.6 (±11.1)	65.3 (±11.4)
	Post	31	63.4 (±12.1)	64.5 (±12.1)
	Follow Up	31	65.8 (±13.8)	67.7 (±6.5)
Control	Pre	41	63.7 (±9.7)	63.9 (±9.1)
	Post	36	67.3 (±9.6)	68.9 (±6.5)
	Follow Up	31	68.6 (±7.0)	68.2 (±8.0)
Treatment Effect				
Treatment effect	Pre–Post		**5.8**	**5.8**
95%HPDI			**1.5–10.0**	**1.8–9.6**
Treatment effect	Pre–Follow Up		**4.7**	1.9
95%HPDI			**0.1–9.2**	–2.3–6.1
Treatment effect	Post–Follow Up		–1.1	–3.9
95%HPDI			–5.7–3.6	–8.2–0.3



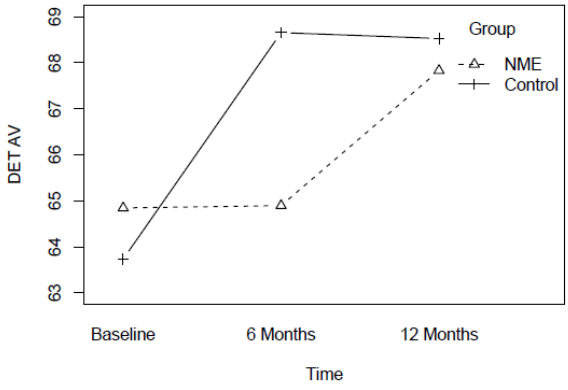


Figure 12 Interaction plots for the pre post and follow up differences between the groups. AD – angular displacement; AV – angular velocity; DET – determinism; NME – neuromuscular exercise. Note: Figure 12 illustrates the empirical means, controlled for the covariates illustrated in Table 9. These can differ slightly from the observed effects.

6 DISCUSSION

A basic component in examining patients with NSLBP is observing lumbopelvic kinematics [7, 9-12]. This observation is a core component of many contemporary classification systems [2, 6-9]. Such examinations are problematic in clinical settings because simple measurement systems lack accuracy, reliability, validity, comprehensiveness, and practicality [13, 14]. A novel wireless IMU system might overcome these limitations. Therefore, the aims of this doctoral thesis were 1) to evaluate the validity and reliability of lumbopelvic kinematics measured with this system and 2) to quantify the effects of NSLBP intensity, fatigue, and NME with valid, reliable indices.

6.1 Reliability and Validity of Lumbopelvic Kinematics Related to NSLBP

The IMU system is concurrently valid to measure lumbopelvic kinematics in the primary movement direction. However, the system appears less valid for assessing movements in non-primary directions. On average, measures of lumbar ROM and movement variability and complexity are more reliable compared to measures of MCI and RE. Furthermore, it was discovered that NSLBP intensity and fatigue affect lumbopelvic kinematics and that NME improves it. The presence of variability and complexity in movement patterns may represent the underlying physiological capability to adapt flexibly to everyday stressors on the neuromuscular system [259, 260]. In contrast, ergonomically poor movement patterns would affect movement variability and complexity, thus elevating tissue loading.

Surprisingly, the association appears to be bidirectional: Painfree people show more deterministic movement than people with NSLBP, but they exhibit more complex and less deterministic movement as a response to fatigue. Six months of NME resulted in preserved, less deterministic movement. Therefore, a nonlinear or U-shaped relationship between movement complexity and variability with disease is possible [261]. This hypothesis is supported by other studies in which people with NSLBP have shown more structured variability of lumbar movement but less structured variability of accessory lumbar movement [223]. People suffering from subacute LBP perform low-

load repetitive work with more variable arm and trunk movements than people suffering from chronic LBP [262]. In summary, while an association exists between movement variability/complexity and NSLBP intensity, fatigue, and NME, the latter's optimal or healthy amount remains unknown.

Our enhanced understanding can help to develop a conceptual framework of lumbar spine function based on these indices (Figure 11). Such a framework could help improve assessment and treatment of people with NSLBP, in whom movement and movement control dysfunctions are regarded as an important clinical characteristic. The following sections illustrate these findings and conclusions in detail.

6.1.1 Concurrent Validity of Lumbopelvic Kinematics (Study I)

Compared to the reference standard, the novel wireless IMU system is a valid alternative for measuring movements in the primary movement direction. However, the system appears less valid for assessing movements in nonprimary directions. Although the RMSE were similar in magnitude compared to the primary movement direction, they were higher relative to the total ROM. Noise, a limited resolution of the IMU system, a nonlinear correlation between both systems, and constraints on mathematical calculations could all affect the validity of the IMU system.

This thesis improves on previous work, as its more detailed analysis of ROM measures includes thoracic spine and hip ROM [17, 263]. Furthermore, the new IMU system compares well to others [17, 263-265]. The ROM measured falls well within the range of previously published results [219]. However, a large variety of measurement approaches hampers comparability, including measurement systems and participant selection [219]. Both our optoelectronic and IMU systems measured similar ROM, whilst sagittal plane movement was slightly overestimated, and frontal plane movement underestimated, by the IMU systems.

6.1.2 Reliability of Lumbopelvic Kinematics (Study II)

On average, the ROM and RM tests needed a smaller number of repeated trials to reach high reliability and had smaller CVs when compared to the MCI and RE tests. This result indicates that within- and between-day variations have less effect on measures of ROM and RM. The complexity and lower standardisation of the MCI and RE tests may decrease their reliability. Segment movement ranges, duration, timing, and speed were not controlled. Standardizing them for one of these factors might decrease within- and

between-day variance. Noise, the IMU system's limited resolution, and constraints on mathematical calculations could affect the reliability of tests with a small ROM. Thus, it might be necessary to apply more accurate measurement tools to investigate tests with small ROM.

The index of dependability Φ of a single trial varied across different tests and variables, ranging from 0.19 to 0.90. The CV varied considerably as well, ranging from <1 to 84%. Increasing the number of trials/days and using the mean value improves reliability. When attempting to increase the number of trials, learning effects and fatigue might influence participants' performances [231]. While averaging over days affected reliability more than averaging over trials on one day for some variables, this solution is not necessarily practical, especially in clinical settings.

6.1.2.1 Reliability of MCI Tests

The MCI tests differed in their reliability. "Waiter's Bow" reached high reliability when measuring one trial on one day. "Sitting Knee Extension" reached high reliability when averaging a maximum of six trials on one day, or two trials on two days with pain-free participants (although it showed poor reliability in people with NSLBP). The magnitude and variability of lumbar spine movement during "Sitting Knee Extension" in the participants with NSLBP were smaller, compared to pain-free participants, possibly explaining the very low Φ. It was assumed that between-subject variance would generally be greater in participants with NSLBP compared to pain-free participants, resulting in greater Φ, but the results vitiate this assumption. The magnitude of the between-day variance, which ranges from 8 to 84%, is also shown by the CV. Nonetheless, the mean ROM in "Sitting Knee Extension" approached zero, with approximately 25% of participants moving into extension, thus hampering the interpretation of the CV (84%) for this variable.

"Pelvic Tilt", "Rocking Forwards", "Rocking Backwards," and "Prone Knee Bend" showed little to moderate reliability. The reliability might have been affected by the complexity or the standardisation of the MCI tests, or because segment movement ranges, timing, duration, and speed were not controlled. Standardizing the MCI tests for one of these factors might decrease within-day and between-days variance. Tests with greater ROM and fixed timing between segments may be less affected by noise, limited resolution of the IMU system, and constraints on mathematical calculations, possibly explaining why those tests are more reliable. Furthermore, between-day effects such as learning effects and changes in pain might have affected test performance and thus reliability. A shorter period between the two measurement sessions might decrease

between-days variance. It is possible that MCI is an unstable phenomenon. Longitudinal studies investigating it might give further insight into whether it is a stable and relevant feature of NSLBP.

Our results are somewhat contradictory in regard to previous research, where the reliability of MCI tests was reported as substantial based on a dichotomous variable (positive or negative indication) [266]. Although a growing body of research investigates MCI of the trunk and hip [258, 266-268], no normative values have been published, aside from this doctoral thesis. Additionally, the different approaches to quantifying MCI tests make it difficult to compare our results.

6.1.2.2 Reliability of RM Tests

The "Picking Up a Box" test had high reliability when averaging a maximum of four trials on one measurement day, with low CVs (≤3%). Our descriptive results for DET of angular displacement are comparable with previous research [223], which did not report on the reliability of their measures.

The "Flexion and Extension" test showed lower Φ-values, whilst the CVs were also small (≤6%). "Picking Up a Box" is predominantly performed by flexing the spine and hips, while the second test is based on flexion and extension. Measures of extension were less reliable and had lower concurrent validity, which might explain the lower Φ values. Biological variability between days and the test setup, where visibility of markers might have been obstructed in end-range positions during full extension, are possible factors compromising reliability and concurrent validity. Both tests were highly standardized, possibly explaining the small standard deviations of these variables.

6.1.3 Associations between NSLBP Intensity and Lumbopelvic Kinematics (Study III)

Greater NSLBP intensity significantly decreased the structure of lumbar movement variability, as shown predominantly in angular velocity and acceleration data. These findings are in line with previous studies investigating gait and repetitive reaching in LBP patients, which related these findings to poor coordination of lumbar erector spinae muscles [269, 270]. To maintain constant movement patterns during repetitive activities, an adaptive muscular activation is necessary, leading to a greater variability of muscular activity and less variability of movement patterns. Variable electromyography

patterns may accompany stereotyped movements, but painful conditions such as LBP may reverse this [271]. Contrary to our findings, one study on patients with chronic NSLBP, using a similar RM test, reported more structured variability — but a different filter might explain this contradiction [223]. Dideriksen and colleagues used a notch filter to smooth out the frequencies related to the RM, preserving other frequency components. This method allowed them to investigate deviations from the target movement, while our interest was the target movement itself. We investigated a guided, standardised target movement, and painfree people might have fewer difficulties learning the tasks and performing them in a deterministic way. The participants were asked to perform the RM tests with a predetermined fast speed, possibly overriding any protective feedforward strategy of movement control. This contrasts with findings from two studies [272, 273], which showed that patients after shoulder surgery had less variable kinematics of the shoulder joint when moving at self-selected slow and fast speeds. To test this hypothesis, the RM tests should be performed at self-selected according to the subject's preference, slow and fast speeds.

In contrast to previous reports, greater NSLBP intensity had no effect on MCI [155]. The lower reliability, as discussed above, of the MCI tests could have influenced the results. Furthermore, methodological differences regarding the group allocation, study population, and measurement systems can explain these differences. Luomajoki and colleagues (2008) grouped patients with chronic LBP and varying degrees of pain intensity together and used a dichotomized, subjective rating [258]. Dichotomizing the participants might have increased the contrast between the two groups. However, using a quantitative approach leads to more detailed insights into the relation between MCI and NSLBP. The specific selection of the study population might have increased the group differences observed. Luomajoki and colleagues (2008) recruited patients with NSLBP undergoing physiotherapy treatment [258]. Conversely, not all participants with NSLBP recruited for this doctoral thesis perceived their condition as serious enough to seek treatment, indicating a lower burden of disease and less impairment due to NSLBP. Luomajoki and colleagues (2008) used a dichotomous rating of movement control by observing lumbar spine flexion, while in the present study the novel wireless IMU system measured MCI continuously and objectively.

The final models show that covariates such as gender and BMI significantly affect nonlinear indices of lumboplevic kinematics (Appendix, Table 12). Their effect was not consistent across all indices, and sometimes exceeded that of NSLBP intensity. Consequently, considering these covariates in future research is recommended. Other covariates possibly relating to indices of lumbopelvic kinematics are the frequency and

duration of the current NSLBP episode and physical stress during leisure time. Furthermore, anthropometric factors, such as a participant's arm length, might impact performance during the "Pick Up a Box" test and should be controlled for in future research. The models and subsequent interpretations were based on the assumption of a linear relationship between perceived NSLBP intensity and lumbopelvic kinematics, which was confirmed by partial-residual plots [238]. Backwards selection of covariates enabled us to test the effect of two-way interactions before testing main effects, and to exclude redundant covariates from the final models. Perceived NSLBP intensity was measured using an NRS and may not have ratio qualities [274]. Therefore, a one-point increase in mild pain intensity may not have the same meaning as it does in high pain intensity. In addition, two participants with similar pain might not rate their pain equally.

The number of participants was unevenly distributed across the levels of perceived NSLBP intensity, with a small number that rated their perceived NSLBP higher than five. However, the distribution of an outcome does not affect the model's validity, if the residuals follow a normal distribution that is verified by residual analysis. Exclusion criteria were based on patient history interviews and questionnaires. To improve the validity of patient selection, ascertainment should be accompanied by anamnestic interviews, physical examination, imaging techniques or other instruments.

This study investigated sagittal plane lumbopelvic kinematics, as this was found to be an important clinical feature in patients with NSLBP [2, 79]. Future studies should expand on this research and address the control of combined movements using a valid measurement system, since NSLBP and injury might occur while combining rotational torques and sagittal or lateral rotations.

In summary, NSLBP intensity affects lumbar movement control. This effect manifests in nonlinear but not in linear indices.

6.1.4 Associations between Fatigue and NSLBP with Lumbopelvic Kinematics (Study IV)

In contrast to participants suffering from NSLBP, painfree participants displayed more complex and less predictable lumbar angular velocity with less structured variability, following an isometric endurance test.

These findings indicate that the presence of NSLBP influences a person's response to fatigue. As a strategy to reduce load on fatiguing tissues while preserving task performance, painfree people might adapt to fatigue by showing more complex and less predictable lumbar angular velocity with a lower degree of structure in its variability.

Evidence suggesting that changes in movement variability may help preserve performance during a fatiguing task has been reported in tests on repetitive throwing, reaching, and elbow flexion/extension for tracking a target, and in cross-country skiing and hammering [275-282]. On the other hand, people suffering from NSLBP may be unable to adapt their movement strategy by exploiting the musculoskeletal system's redundancy, thus accumulating load on fatiguing tissues. Pain-induced changes in muscle and motor unit recruitment patterns could explain this inability, as indicated by studies using experimentally induced pain [283-286].

Contrariwise, the present findings could indicate that the painfree participants were able to control their lumbar movement until the onset of fatigue, thereby resembling participants suffering from NSLBP. The less complex and more predictable lumbar angular velocity with more structured variability shown by painfree participants during the pretest could point to that. This factor would imply that more structured movement variability might actually be beneficial and represent better movement control, supporting the findings of Study III. Furthermore, it might indicate an inability of the lumbar paraspinal muscles to stabilize the lumbar spine in the neutral zone. A study on lumbar muscle recruitment patterns would address this hypothesis.

The presence of variability in movement patterns may represent the underlying physiological capability to make flexible adaptations to everyday stressors placed on the neuromuscular system [259, 260]. In contrast, the presence of ergonomically poor movement patterns might increase the movement variability measured, thus elevating tissue loading. Therefore, a non-linear or U-shaped relationship between structure, complexity and disease is hypothesised [261], as reported by previous studies. More structured variability of lumbar movement (Study III) but less structured variability of acessory lumbar movement [223] were associated with LBP, which might indicate early functional manifestations of LBP. Larger and smaller sizes of arm and trunk movement variability were found during simulated low-load repetitive work in people suffering from acute and sub-acute or chronic pain, respectively [262].

A limitation of the present study is that the true size of the effect of fatigue on the complexity of lumbar movement and the degree of structure in its variability remain unknown, since this study did not include a measure of lumbar muscular fatigue. In future, EMG data should be recorded to supplement kinematic analyses, and changes in DET of EMG data should be analysed both pre- and post-fatigue. Increases in DET have been reported in the presence of fatigue, which reflects higher periodicity [287]. The inclusion of a measure of lumbar strength could have helped quantify if both groups were equally influenced by the protocol. Future studies might consider additional lateral flexion and rotation angles. They were not analysed, due to the IMU

systems' limited concurrent validity when measuring lateral flexion or rotation movements of small magnitude during large flexion extension movements. Furthermore, the study design is not set up to determine if the difference in response to fatigue is a cause or a result of NSLBP. Still, this doctoral thesis demonstrated that changes in lumbar movement velocity after an isometric endurance test are influenced by the presence of NSLBP.

In summary, the presence of NSLBP influences a participant's response to an isometric endurance test. This response manifests in nonlinear indices of lumbar angular velocity.

6.1.5 Effect of Neuromuscular Exercise on Lumbopelvic Kinematics (Study V)

NME decreases the structure of lumbar movement variability. The observed treatment effect is substantial, as its magnitude is approximately one-half of the standard deviations of the outcomes **Table 9**. The structure of lumbar movement variability increased in a control group, which received no intervention for their NSLBP, compared to the NME group. Thus, NME improves or maintains neuromuscular functional integrity, indicating an increased capacity of the neuromuscular system to generate adequate responses to stressors and to function during demanding tasks.

Sufficient lumbar motor variability ensures that new movement solutions can be adopted in response to changes such as sudden perturbations, and may therefore be relevant for maintaining occupational health and performance [179, 282]. People with chronic LBP show less variability in trunk and lumbar movement during gait and repetitive lifting [223, 288]. People with chronic pain may prefer more stereotypical motor solutions to possible alternatives, although those might lead to faster fatiguing of trunk muscles, decreased task performance, and prolonged stereotypical loading of the painful area [277, 289, 290]. Our results indicate that the structure of lumbar movement variability may decrease further when untreated, demonstrating an inability of the neuromuscular system to recapture its integrity, whereas NME may reverse or reduce this pain-related complexity loss of the neuromuscular system [282, 291, 292].

Motor variability increases in the short term following task-specific training such as biofeedback training for office workers [293], and it increases in the long term due to skill development during repetitive occupational tasks such as lifting [294]. While these previous studies investigated the short-term effect of training and the natural course of skill development, our results indicate that NME increases movement variability in the

NME group, compared to the control group, over a six-month period in people suffering from subacute NSLBP. The treatment effect diminished between 6 and 12 months of follow up, indicating that NME training that is prolonged, intensive, and guided might be necessary to maintain movement variability in a high-risk population such as nurses. The optimal NME design for achieving long-term improvement of lumbar movement variability of greater magnitude remains unknown (e.g., one that would address factors such as training dosage, intensity, or type of feedback). However, this doctoral thesis is the first to demonstrate that NME can improve lumbar movement variability, or impede further deterioration, over six months' duration.

In summary, NME improves lumbar movement variability, or impedes further deterioration, over 6 months' duration. The optimal NME design (e.g., training dosage, intensity, or type of feedback) to achieve long-term improvement remains unknown.

6.2 Methods and Subjects

6.2.1 Methodological Considerations

While it is tempting to move beyond visual observation, this thesis highlights certain limitations researchers and clinicians might encounter when doing so. The validity of novel measurement systems cannot always be assumed, as illustrated by the limited validity of this new IMU system for measurements in secondary movement directions. Future studies might consider smaller measurement systems and mathematical options that increase concurrent validity. Movement tests might not always be reliable when analysed with an objective, quantitative measurement system, even if they are reliable based on visual observation and dichotomous ratings. These findings might convince researchers and clinicians alike to reconsider theories and treatment approaches based on less objective measurement systems. Contrariwise, a number of decisions an investigator has to make while obtaining and processing data influences the results: because choosing an appropriate filtering technique is a compromise between loss of information and noise allowed through, relevant information from higher frequency contents was possibly lost. Future studies should address options that might conserve such information. When using nonlinear methods such as RQA, the number of decisions increases [295]. This factor limits the comparability of results from different

research projects, especially when complex, nonconvertible methods of data analysis are applied. The participants with NSLBP reported an average mean intensity of three points on the NRS, or mild NSLBP. In future work on lumbar spine function, it might be worthwhile to focus on patients with a wider spectrum of complaints including such with higher intensity NSLBP and a greater burden of disease. A limitation of Study IV is that the true size of the effect of fatigue and its influence on both groups remains unknown, since this study did not measure lumbar muscular fatigue using electromyography or lumbar strength assessments. Study V was planned independently from studies I-IV resulting in different pain intensity tests used. However, there are only small differences in the relative validity of both instruments which were therefore considered equal measures for the purpose of this study [296]. The large number of dropouts in Study V might have decreased the precision of the treatment effect estimation, although the nurses' shift work can partly explain this pattern. Nonetheless, it is important to discover preventive interventions that hinder the development of chronic, more disabling NSLBP in working populations with subacute NSLBP. An assumption of a linear relationship between the quantities of interest was the basis of all statistical analyses herein. Thus, future studies should explore possible nonlinear relations, such as the hypothised U -shaped relationship.

6.2.2 Subjects

This thesis investigated an important patient population that accounts for a large number of visits to health care providers, since LBP is a major international health problem that poses substantial challenges for clinical management and health services [1, 24-26]. Most participants with NSLBP recruited for this thesis presented low to moderate intensity of NSLBP. This patient population is frequently encountered in daily practice, and recruited for a large number of studies on movement and movement-control impairments [28, 54, 57, 94, 99, 102, 104].

Identifying pain-driving factors in patients with NSLBP remains a major research challenge, and because NSLBP is a multidimensional problem, not all patients show impairments of movement or movement control [4, 155]. Contrariwise, if both NSLBP patients and painfree people showed similar lumbopelvic kinematics, the clinical relevance of the latter would be doubtful. The patients with NSLBP in this thesis were not screened for impairments in movement or movement control: Establishing first knowledge on the validity and reliably of this novel measurement system required a broad spectrum of subjects, with and without movement and movement control

impairments. Although the first evaluation of a novel measurement system cannot be translated into diagnostic action, it adds to our biological insight into the impairment and may serve later research into diagnosis and treatment [155, 297].

Given their unfavourable prognosis of patients with recurrent episodes of NSLBP, future research on the diagnostic value of lumbopelvic kinematics should focus on them [59]. Deteriorated lumbar movement variability and complexity are candidates for early markers of recurrent NSLBP, and might respond to treatments such as NME.

This thesis provides a large body of normative values for indices of lumbopelvic kinematics, for both painfree individuals and those suffering from NSLBP. Normative values are important for the interpretation of assessments and treatment effects. Certainly, normative values are population specific, so they apply to the populations assessed in this thesis. Furthermore, they are difficult to collect.

As stated in the literature review and methods sections, different indices of lumbopelvic kinematics exist. Since the novel IMU system provides valid, reliable estimates for some but not all of them, it is useful for specific patients. Factors such as movement direction, ROM, and test standardization are important when determining if the novel IMU system is useful for assessing an individual patient.

6.2.3 Clinical Value

The literature describes a variety of classification systems for patients experiencing LBP which divide peripherally mediated pain disorders of the lumbar spine into those involving impairment either in movement or movement control [2, 6-9, 155]. However, interventions aiming to address either rarely apply objective, quantifying measures, often relying on visual observation of patients when classifying their condition [98]. The novel, wireless IMU system can be used to quantify lumbopelvic kinematics and thus to validate these classification systems and therapy approaches. Their underlying hypothesis, that both impairments are a clinically relevant factor in the assessment and treatment of NSLBP, can be addressed by objectively investigating factors such as their prevalence, incidence, responsiveness to treatment, and association with clinically relevant outcomes. Previously, the examination of lumbopelvic kinematics was problematic because simple measurement systems such as visual observation lacked accuracy, reliability, and validity, while reference standards were not applicable due to issues such as their high cost [13, 14]. The novel IMU system overcomes these limitations and enables clinicians to assess their patients with NSLBP objectively and

researchers to conduct studies on the clinical relevance of impairments in movement and movement control. Using the novel IMU system, clinicians can assess their patients' movements and movement control, monitor their treatment effects, and thus test their own hypotheses on their patients' underlying problems. Every measurement system has a measurement error. Whether that error is small enough in a clinical context depends on the proposed use, with the degree of acceptable measurement error relating directly to the intended application [298]. On the basis of our data, assessing movements with a large amplitude and a primary movement direction can be recommended for a clinical context. Movements with small amplitudes into a secondary movement direction should be interpreted with caution.

The novel IMU system is the core of a technology-supported clinical therapy system (Valedo®, Hocoma AG, Volketswil, Switzerland). Generally, technology-supported therapy appears to improve pain, disability, and quality of life for patients with chronic NSLBP. When added to conventional treatment, technology-supported treatment is superior to conventional treatment alone. However, technology-supported therapy alone is not superior to other interventions [299]. One explanation for this lack of benefit might be the narrow approach of most technology-supported treatments (e.g., providing training for only one particular function or movement) [299]. Since the novel IMU system provides valid, reliable measures of various lumbopelvic kinematics, it offers a larger selection of assessments and movement therapies, overcoming this limitation. The clinical therapy system individualizes treatment approaches seeking to restore movement and movement control. Individualizing therapy for both with respect to individual skill level, demands, preferences, and potential moderating factors such as pain or physical activity may lead to greater therapy success [300, 301]. However, the optimal dose and setting of series game exercises are still unknown. Together with a user interface and serious games, the clinical therapy system offers clinicians and patients innovative features such as visual feedback [299, 302]. Attempts to use visual feedback to reinstate the normal precision of bodily representations and normal movement in LBP are still in their infancy, but early results seem promising [302, 303]. It might be promising to integrate visual feedback based on IMU systems input into conventional therapy for LBP. The clinical therapy system comprises home-exercise programs, but 50–70% of patients with chronic LBP lack adherence to such programs in general [304, 305]. Improvement seems warranted, since good adherence is a predictor for treatment success [306]. The novel clinical therapy system might promote adherence, as indicated in research on populations with diabetes and obesity [307, 308]. The author's study group investigated adherence to a home exercise program based on an early version of the novel clinical therapy system, and found it was not superior to

conventional home exercises [309]. This result suggests that solely providing patients with technological support at home does not increase adherence. Instead, future versions of the novel therapy system should follow a user-centred design approach [310]. To ensure that user needs remain at the forefront of development, such an approach employs design ethnography and participatory stakeholder involvement as key drivers for technology development. It also uses an iterative product design methodology, which re-evaluates and improves the user appropriateness of the system at each stage [311]. Key user groups for the novel clinical therapy system are patients and clinicians. Following this process very well might advance the clinical therapy system so that it could fulfil its potential to improve patients' adherence to home exercises.

6.3 Outlook

Comparing lumbopelvic kinematics with muscle recruitment patterns and work performance measures will help to explore the hypothesised U-shaped relationship between lumbopelvic kinematics and NSLBP. It will improve our understanding of the underlying mechanisms by evaluating whether the observed changes in lumbar movement variability under pain and following fatigue preserve an individual's task performance. Longitudinal prospective studies should investigate the development of NSLBP and corresponding changes in lumbar movement variability, to assess whether changes in it are a cause or a consequence of NSLBP.

The conceptual framework of lumbar spine function should be investigated for its predictive, content, and construct validity, as well as responsiveness [227]. This process will require investigations on the effect of other possibly relevant items (e.g., the anthropometric factors of age, gender, and BMI, or other factors such as yellow flags), and on the question of whether the concept of lumbar spine function has reflective or formative properties [227] (Figure 13).

The IMU can quantify lumbopelvic kinematics objectively. Due to its low cost and spatial flexibility, data acquisition can take place outside of the lab, giving researchers and clinicians more options. Investigating and subsequently improving the technical validity of its components (for instance, regarding their size) might enhance the IMU system's technical validity. Thus, future studies might be able to consider multidirectional movements. One general limitation of IMU systems is their inability to measure translational movements, in addition to angular rotations. In a follow-up project, the author and his study group have begun to integrate the IMU system with

an infrared optical measurement system. With the ability to measure both movement components, this integrated system will predominantly target patients with neck pain. The aforementioned commercialisation of the IMU system and the continuing interest of funding organisations indicate the clinical, scientific, and economical potential of this technology. To identify dysfunctions and changes in performance over time, high reliability is important. Future studies should address reliability in different populations and assess in more detail the IMU system's diagnostic value and ability to detect changes of the presented indices over time. Regarding factors such as training dosage, intensity, or type of feedback, the optimal NME design necessary for long-term improvement in lumbar movement variability is subject to a future line of research. Nonlinear indices depend greatly on input parameters [242, 295, 312, 313]. Other choices for these parameters are possible, apart from those used in this thesis (Table 11), and they should be investigated in future studies.

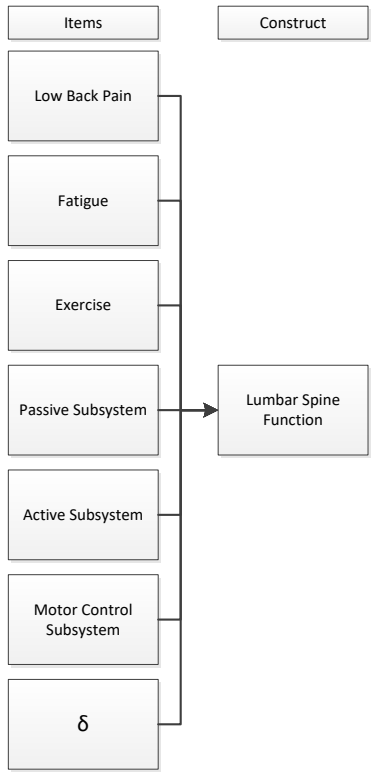


Figure 13. Formative conceptual framework of lumbar spine function, based on the three subsystems maintaining the spine within physiological limits and the factors investigated in this doctoral thesis. This conceptual framework could help to improve assessment and treatment of

people with NSLBP where functional pathology of the lumbar spine is regarded an important clinical characteristic. Based on [227] and [314]. δ = Error term, the unobserved factors influencing lumbar spine function not yet investigated and therefore not taken into account for the framework

7 CONCLUSION

The IMU system is concurrently valid to measure lumbopelvic kinematics in the primary movement direction. However, the system appears less valid for assessing movements in nonprimary directions.

On average, the ROM and RM tests needed a smaller number of repeated trials to reach high reliability and had smaller CVs when compared to the MCI and RE tests. This result indicates that within- and between-day variations have less effect on measures of ROM and RM.

Because NSLBP intensity affects lumbopelvic kinematics, people with NSLBP show more recurrent and predictable lumbar movement.

Fatigue also affects lumbopelvic kinematics. This effect depends on the presence of NSLBP. Painfree people might adjust to fatigue by using a strategy to reduce load on fatiguing tissues while preserving task performance. Contrariwise, the present findings could indicate that the painfree participants were able to control their lumbar movement but lost control after the onset of fatigue, thereupon resembling people suffering from NSLBP.

NME affects lumbopelvic kinematics. When compared to no intervention, it may reverse or reduce deterioration of lumbar movement by decreasing or preserving the degree of structure of the movement's variability.

This thesis raises several questions regarding improvements to the IMU system, the optimal or healthy amount of lumbar movement complexity and variability, the optimal NME design, and more. The authors study group is currently working to improve the IMU system so that it will be able to measure both angular rotation and translation. The continuing interest of funding organisations and commercial partners indicates the clinical, scientific, and economical potential of this conceptual framework and its underlying technology.

8 SUMMARY

A basic component in assessing patients with NSLBP is examining lumbopelvic kinematics. This examination is problematic because simple measurement systems such as visual observation or goniometers lack accuracy, reliability, validity, comprehensiveness, and practicality. To overcome these limitations, this doctoral thesis introduces a novel, wireless, movement-analysis system based on IMUs. It focuses on the concurrent validity of the novel IMU system and the reliability of lumbopelvic kinematics it measures. Through using this system, the effects of NSLBP intensity, fatigue, and exercise therapy on lumbopelvic kinematics are explored.

The IMU system is concurrently valid to measure lumbopelvic kinematics in the primary movement direction. However, it appears less so for assessing movements in nonprimary directions. On average, measures of lumbar ROM and movement variability and complexity are more reliable, compared to measures of movement-control impairments and reposition error.

Furthermore, it was discovered that NSLBP intensity and fatigue affect lumbopelvic kinematics and that exercise therapy improves them. Because NSLBP intensity affects lumbopelvic kinematics, people with NSLBP show more recurrent and more predictable lumbar movement. Depending on the presence of NSLBP, fatigue also affects them. Painfree people might adjust to fatigue by using a strategy to reduce load on fatiguing tissues while preserving task performance. Exercise therapy affects lumbopelvic kinematics, and when compared to no intervention it may reverse or reduce deterioration of lumbar movement, by increasing or preserving movement variability.

9 REFERENCES

1. Costa-Black, K.M., et al., *Back pain and work.* Best Pract Res Clin Rheumatol, 2010. **24**(2): p. 227-40.

2. O'Sullivan, P., *Diagnosis and classification of chronic low back pain disorders: Maladaptive movement and motor control impairments as underlying mechanism.* Manual Therapy, 2005. **10**(4): p. 242-255.

3. Delitto, A., et al., *Low back pain.* J Orthop Sports Phys Ther, 2012. **42**(4): p. A1-57.

4. Airaksinen, O., et al., *Chapter 4. European guidelines for the management of chronic nonspecific low back pain.* Eur Spine J, 2006. **15 Suppl 2**: p. S192-300.

5. Borkan, J.M., et al., *A report from the Second International Forum for Primary Care Research on Low Back Pain. Reexamining priorities.* Spine, 1998. **23**(18): p. 1992-6.

6. Delitto, A., R.E. Erhard, and R.W. Bowling, *A treatment-based classification approach to low back syndrome: identifying and staging patients for conservative treatment.* Phys Ther, 1995. **75**(6): p. 470-85; discussion 485-9.

7. McKenzie, R. and S. May, *The lumbar spine: mechanical diagnosis and therapy*. Vol. 1. 2003: Orthopedic Physical Therapy.

8. Petersen, T., et al., *Diagnostic classification of non-specific low back pain. A new system integrating patho-anatomic and clinical categories.* Physiotherapy Theory and Practice, 2003. **19**(4): p. 213-237.

9. Sahrmann, S., *Diagnosis and treatment of movement impairment syndromes*. 2002: Elsevier Health Sciences.

10. Liebenson, C., *Rehabilitation of the spine: a practitioner's manual*. 2007: Lippincott Williams & Wilkins.

11. Magee, D.J., *Orthopedic physical assessment*. 2014: Elsevier Health Sciences.

12. Maitland, G.D., et al., *Maitland's vertebral manipulation*. Vol. 1. 2005: Butterworth-Heinemann.

13. Taulaniemi, R., et al., *Reliability of Musculoskeletal Fitness Tests and Movement Control Impairment Test Battery in Female Health-Care Personnel with Re-Current Low Back Pain.* Journal of Novel Physiotherapies, 2016. **6**(1).

14. Chen, S.P., et al., *Reliability of three lumbar sagittal motion measurement methods: surface inclinometers.* J Occup Environ Med, 1997. **39**(3): p. 217-23.

15. Cuesta-Vargas, A.I., A. Galan-Mercant, and J.M. Williams, *The use of inertial sensors system for human motion analysis.* Phys Ther Rev, 2010. **15**(6): p. 462-473.

16. McGinley, J.L., et al., *The reliability of three-dimensional kinematic gait measurements: a systematic review.* Gait Posture, 2009. **29**(3): p. 360-9.

17. Wong, W.Y. and M.S. Wong, *Measurement of Postural Change in Trunk Movements Using Three Sensor Modules.* Ieee Transactions on Instrumentation and Measurement, 2009. **58**(8): p. 2737-2742.

18. Balagué, F., et al., *Non-specific low back pain.* The Lancet, 2012. **379**(9814): p. 482-491.

19. Hoy, D., et al., *The Epidemiology of low back pain.* Best Pract Res Clin Rheumatol, 2010. **24**(6): p. 769-81.

20. Minematsu, A., *Epidemiology*, in *Low Back Pain*, A.A. Norasteh, Editor. 2012, InTech: Rijeka, Croatia.

21. Hiyama, A., et al., *Evaluation of quality of life and neuropathic pain in patients with low back pain using the Japanese Orthopedic Association Back Pain Evaluation Questionnaire.* European Spine Journal, 2015. **24**(3): p. 503-512.

22. World Health Organization, *Priority Medicines for Europe and the World Update Report, 2013* in *Medicines and Health Products*, WHO, Editor. 2013, WHO.

23. Wieser, S., et al., *Cost of low back pain in Switzerland in 2005.* Eur J Health Econ, 2011. **12**(5): p. 455-67.

24. Kent, P.M. and J.L. Keating, *The epidemiology of low back pain in primary care.* Chiropr Osteopat, 2005. **13**: p. 13.

25. Steenstra, I.A., et al., *Prognostic factors for duration of sick leave in patients sick listed with acute low back pain: a systematic review of the literature.* Occup Environ Med, 2005. **62**(12): p. 851-60.

26. Thelin, A., S. Holmberg, and N. Thelin, *Functioning in neck and low back pain from a 12-year perspective: a prospective population-based study.* J Rehabil Med, 2008. **40**(7): p. 555-61.

27. McAlindon, T.E., et al., *OARSI guidelines for the non-surgical management of knee osteoarthritis.* Osteoarthritis and Cartilage, 2014. **22**(3): p. 363-388.

28. Stanton, T.R., et al., *After an episode of acute low back pain, recurrence is unpredictable and not as common as previously thought.* Spine, 2008. **33**(26): p. 2923-8.

29. Wasiak, R., et al., *Recurrence of low back pain: definition-sensitivity analysis using administrative data.* Spine, 2003. **28**(19): p. 2283-91.

30. Battie, M.C. and T. Videman, *Lumbar disc degeneration: epidemiology and genetics.* J Bone Joint Surg Am, 2006. **88 Suppl 2**: p. 3-9.

31. Boden, S.D., et al., *Abnormal magnetic-resonance scans of the lumbar spine in asymptomatic subjects. A prospective investigation.* J Bone Joint Surg Am, 1990. **72**(3): p. 403-8.

32. Hitselberger, W.E. and R.M. Witten, *Abnormal myelograms in asymptomatic patients.* J Neurosurg, 1968. **28**(3): p. 204-6.

33. Wiesel, S.W., et al., *A study of computer-assisted tomography. I. The incidence of positive CAT scans in an asymptomatic group of patients.* Spine, 1984. **9**(6): p. 549-51.

34. Bener, A., et al., *Obesity and low back pain.* Coll Antropol, 2003. **27**(1): p. 95-104.

35. Picavet, H.S. and J.S. Schouten, *Musculoskeletal pain in the Netherlands: prevalences, consequences and risk groups, the DMC(3)-study.* Pain, 2003. **102**(1-2): p. 167-78.

36. Picavet, H.S., J.S. Schouten, and H.A. Smit, *Prevalence and consequences of low back problems in The Netherlands, working vs non-working population, the MORGEN-Study. Monitoring Project on Risk Factors for Chronic Disease.* Public Health, 1999. **113**(2): p. 73-7.

37. Santos-Eggimann, B., et al., *One-year prevalence of low back pain in two Swiss regions: estimates from the population participating in the 1992-1993 MONICA project.* Spine, 2000. **25**(19): p. 2473-9.

38. Dionne, C.E., K.M. Dunn, and P.R. Croft, *Does back pain prevalence really decrease with increasing age? A systematic review.* Age Ageing, 2006. **35**(3): p. 229-34.

39. Lawrence, R.C., et al., *Estimates of the prevalence of arthritis and selected musculoskeletal disorders in the United States.* Arthritis Rheum, 1998. **41**(5): p. 778-99.

40. Loney, P.L. and P.W. Stratford, *The prevalence of low back pain in adults: a methodological review of the literature.* Phys Ther, 1999. **79**(4): p. 384-96.

41. Dionne, C.E., et al., *Formal education and back pain: a review.* J Epidemiol Community Health, 2001. **55**(7): p. 455-68.

42. Reisbord, L.S. and S. Greenland, *Factors associated with self-reported back-pain prevalence: a population-based study.* J Chronic Dis, 1985. **38**(8): p. 691-702.

43. Matsui, H., et al., *Risk indicators of low back pain among workers in Japan. Association of familial and physical factors with low back pain.* Spine, 1997. **22**(11): p. 1242-7; discussion 1248.

44. Waters, T., et al., *The impact of operating heavy equipment vehicles on lower back disorders.* Ergonomics, 2008. **51**(5): p. 602-36.

45. Hamberg-van Reenen, H.H., et al., *A systematic review of the relation between physical capacity and future low back and neck/shoulder pain.* Pain, 2007. **130**(1-2): p. 93-107.

46. George, S.Z., J.M. Fritz, and J.D. Childs, *Investigation of elevated fear-avoidance beliefs for patients with low back pain: a secondary analysis involving patients enrolled in physical therapy clinical trials.* J Orthop Sports Phys Ther, 2008. **38**(2): p. 50-8.

47. George, S.Z., J.M. Fritz, and D.W. McNeil, *Fear-avoidance beliefs as measured by the fear-avoidance beliefs questionnaire: change in fear-avoidance beliefs questionnaire is predictive of change in self-report of disability and pain intensity for patients with acute low back pain.* Clin J Pain, 2006. **22**(2): p. 197-203.

48. Pincus, T., et al., *A systematic review of psychological factors as predictors of chronicity/disability in prospective cohorts of low back pain.* Spine, 2002. **27**(5): p. E109-20.

49. Wessels, T., et al., *What predicts outcome in non-operative treatments of chronic low back pain? A systematic review.* Eur Spine J, 2006. **15**(11): p. 1633-44.

50. McIntosh, G., H. Hall, and C. Boyle, *Contribution of nonspinal comorbidity to low back pain outcomes.* Clin J Pain, 2006. **22**(9): p. 765-9.

51. van der Hulst, M., M.M. Vollenbroek-Hutten, and M.J. Ijzerman, *A systematic review of sociodemographic, physical, and psychological predictors of multidisciplinary rehabilitation-or, back school treatment outcome in patients with chronic low back pain.* Spine, 2005. **30**(7): p. 813-25.

52. Van Tulder, M., et al., *Chapter 3 European guidelines for the management of acute nonspecific low back pain in primary care.* European spine journal, 2006. **15**: p. s169-s191.

53. Waddell, G., et al., *Clinical guidelines for the management of acute low back pain: low back pain evidence review.* London: Royal College of General Practitioners, 1996.

54. Henschke, N., et al., *Prognosis in patients with recent onset low back pain in Australian primary care: inception cohort study.* Bmj, 2008. **337**: p. a171.

55. Maher, C., M. Underwood, and R. Buchbinder, *Non-specific low back pain.* The Lancet, 2017. **389**(10070): p. 736-747.

56. Menezes Costa, L.d.C., et al., *The prognosis of acute and persistent low-back pain: a meta-analysis.* CMAJ : Canadian Medical Association Journal, 2012. **184**(11): p. E613-E624.

57. Costa, L.d.C.M., et al., *Prognosis for patients with chronic low back pain: inception cohort study.* Bmj, 2009. **339**: p. b3829.

58. Bergquist-Ullman, M. and U. Larsson, *Acute low back pain in industry: a controlled prospective study with special reference to therapy and confounding factors.* Acta Orthopaedica Scandinavica, 1977. **48**(sup170): p. 1-117.

59. Von Korff, M., et al., *Back pain in primary care. Outcomes at 1 year.* Spine (Phila Pa 1976), 1993. **18**(7): p. 855-62.

60. Kuslich, S.D., C.L. Ulstrom, and C.J. Michael, *The tissue origin of low back pain and sciatica: a report of pain response to tissue stimulation during operations on the lumbar spine using local anesthesia.* Orthop Clin North Am, 1991. **22**(2): p. 181-7.

61. Panjabi, M.M., *The stabilizing system of the spine. Part II. Neutral zone and instability hypothesis.* Journal of spinal disorders & techniques, 1992. **5**(4): p. 390-397.

62. DePalma, M.J., J.M. Ketchum, and T. Saullo, *What is the source of chronic low back pain and does age play a role?* Pain medicine, 2011. **12**(2): p. 224-233.

63. Savage, R.A., G.H. Whitehouse, and N. Roberts, *The relationship between the magnetic resonance imaging appearance of the lumbar spine and low back pain, age and occupation in males.* Eur Spine J, 1997. **6**(2): p. 106-14.

64. Boos, N., et al., *Natural history of individuals with asymptomatic disc abnormalities in magnetic resonance imaging: predictors of low back pain-related medical consultation and work incapacity.* Spine, 2000. **25**(12): p. 1484-92.

65. Von Korff, M., et al., *Effects of practice style in managing back pain.* Ann Intern Med, 1994. **121**(3): p. 187-95.

66. Von Korff, M. and K. Saunders, *The course of back pain in primary care.* Spine (Phila Pa 1976), 1996. **21**(24): p. 2833-7; discussion 2838-9.

67. *Scientific approach to the assessment and management of activity-related spinal disorders. A monograph for clinicians. Report of the Quebec Task Force on Spinal Disorders.* Spine, 1987. **12**(7 Suppl): p. S1-59.

68. Karayannis, N.V., G.A. Jull, and P.W. Hodges, *Movement-based subgrouping in low back pain: synergy and divergence in approaches.* Physiotherapy, 2016. **102**(2): p. 159-69.

69. Widerstrom, B., et al., *Feasibility of the subgroup criteria included in the treatment-strategy-based (TREST) classification system (CS) for patients with non-specific low back pain (NSLBP).* Man Ther, 2016. **23**: p. 90-7.

70. Brennan, G.P., et al., *Identifying subgroups of patients with acute/subacute "nonspecific" low back pain: results of a randomized clinical trial.* Spine, 2006. **31**(6): p. 623-31.

71. Childs, J.D., et al., *A clinical prediction rule to identify patients with low back pain most likely to benefit from spinal manipulation: a validation study.* Ann Intern Med, 2004. **141**(12): p. 920-8.

72. Dankaerts, W., et al., *Differences in sitting postures are associated with nonspecific chronic low back pain disorders when patients are subclassified.* Spine, 2006. **31**(6): p. 698-704.

73. Harris-Hayes, M. and L.R. Van Dillen, *The inter-tester reliability of physical therapists classifying low back pain problems based on the movement system impairment classification system.* Pm r, 2009. **1**(2): p. 117-26.

74. Hicks, G.E., et al., *Preliminary development of a clinical prediction rule for determining which patients with low back pain will respond to a stabilization exercise program.* Arch Phys Med Rehabil, 2005. **86**(9): p. 1753-62.

75. Long, A., R. Donelson, and T. Fung, *Does it matter which exercise? A randomized control trial of exercise for low back pain.* Spine, 2004. **29**(23): p. 2593-602.

76. Rabin, A., et al., *A clinical prediction rule to identify patients with low back pain who are likely to experience short-term success following lumbar stabilization exercises: a randomized controlled validation study.* J Orthop Sports Phys Ther, 2014. **44**(1): p. 6-b13.

77. Trudelle-Jackson, E., S.A. Sarvaiya-Shah, and S.S. Wang, *Interrater reliability of a movement impairment-based classification system for lumbar spine syndromes in patients with chronic low back pain.* J Orthop Sports Phys Ther, 2008. **38**(6): p. 371-6.

78. Vibe Fersum, K., *Classification and targeted treatment of patients with non specific chronic low back pain.* 2011.

79. Vibe Fersum, K., et al., *Inter-examiner reliability of a classification system for patients with non-specific low back pain.* Manual Therapy, 2009. **14**(5): p. 555-561.

80. Dankaerts, W., et al., *The inter-examiner reliability of a classification method for non-specific chronic low back pain patients with motor control impairment.* Man Ther, 2006. **11**(1): p. 28-39.

81. Bombardier, C., *Outcome assessments in the evaluation of treatment of spinal disorders: summary and general recommendations.* Spine, 2000. **25**(24): p. 3100-3.

82. Deyo, R.A., et al., *Outcome measures for low back pain research. A proposal for standardized use.* Spine, 1998. **23**(18): p. 2003-13.

83. Gatchel, R.J., et al., *Treatment- and cost-effectiveness of early intervention for acute low-back pain patients: a one-year prospective study.* J Occup Rehabil, 2003. **13**(1): p. 1-9.

84. Gellhorn, A.C., et al., *Management patterns in acute low back pain: the role of physical therapy.* Spine, 2012. **37**(9): p. 775-82.

85. Hagen, E.M., H.R. Eriksen, and H. Ursin, *Does early intervention with a light mobilization program reduce long-term sick leave for low back pain?* Spine, 2000. **25**(15): p. 1973-6.

86. Linton, S.J., A.L. Hellsing, and D. Andersson, *A controlled study of the effects of an early intervention on acute musculoskeletal pain problems.* Pain, 1993. **54**(3): p. 353-9.

87. Nordeman, L., et al., *Early access to physical therapy treatment for subacute low back pain in primary health care: a prospective randomized clinical trial.* Clin J Pain, 2006. **22**(6): p. 505-11.

88. Pinnington, M.A., J. Miller, and I. Stanley, *An evaluation of prompt access to physiotherapy in the management of low back pain in primary care.* Fam Pract, 2004. **21**(4): p. 372-80.

89. Wand, B.M., et al., *Early intervention for the management of acute low back pain: a single-blind randomized controlled trial of biopsychosocial education, manual therapy, and exercise.* Spine, 2004. **29**(21): p. 2350-6.

90. Kovacs, F.M., et al., *The transition from acute to subacute and chronic low back pain: a study based on determinants of quality of life and prediction of chronic disability*. Spine, 2005. **30**(15): p. 1786-1792.

91. Lethola, V., *Movement Control Impairment in Recurrent Subacute Low Back Pain - A Randomized Controlled Trial Between Specific Movement Control Exercises and General Exercises*. 2017, University of Eastern Finland.

92. Hayden, J., et al., *Exercise therapy for treatment of non-specific low back pain.* The Cochrane Library, 2005.

93. Ebadi, S., et al., *Therapeutic ultrasound for chronic low-back pain.* Cochrane Database Syst Rev, 2014(3): p. Cd009169.

94. Franke, H., et al., *Muscle energy technique for non-specific low-back pain.* Cochrane Database Syst Rev, 2015(2): p. Cd009852.

95. Furlan, A.D., et al., *Massage for low-back pain.* Cochrane Database Syst Rev, 2015(9): p. Cd001929.

96. Kalin, S., A.K. Rausch-Osthoff, and C.M. Bauer, *What is the effect of sensory discrimination training on chronic low back pain? A systematic review.* BMC Musculoskelet Disord, 2016. **17**(1): p. 143.

97. Kamper, S.J., et al., *Multidisciplinary biopsychosocial rehabilitation for chronic low back pain.* The Cochrane Library, 2014.

98. Luomajoki, H.A., et al., *Effectiveness of movement control exercise on patients with non-specific low back pain and movement control impairment: A systematic review and meta-analysis.* Musculoskelet Sci Pract, 2018. **36**: p. 1-11.

99. Macedo, L.G., et al., *Motor control exercise for acute non-specific low back pain.* The Cochrane Library, 2016.

100. Poquet, N., et al., *Back schools for acute and subacute non-specific low-back pain.* Cochrane Database Syst Rev, 2016. **4**: p. Cd008325.

101. Rubinstein, S.M., et al., *Spinal manipulative therapy for acute low back pain: an update of the cochrane review.* Spine, 2013. **38**(3): p. E158-77.

102. Saragiotto, B.T., et al., *Motor control exercise for chronic non-specific low-back pain.* The Cochrane Library, 2016.

103. Wegner, I., et al., *Traction for low-back pain with or without sciatica.* Cochrane Database Syst Rev, 2013(8): p. Cd003010.

104. Yamato, T.P., et al., *Pilates for Low Back Pain: Complete Republication of a Cochrane Review.* Spine, 2016. **41**(12): p. 1013-21.

105. Vickers, A. and C. Zollman, *ABC of complementary medicine: Massage therapies.* British Medical Journal, 1999. **319**(7219): p. 1254.

106. Brooks, C., S. Kennedy, and P.W. Marshall, *Specific trunk and general exercise elicit similar changes in anticipatory postural adjustments in patients with chronic low back pain: a randomized controlled trial.* Spine, 2012. **37**(25): p. E1543-50.

107. da Fonseca, J.L., M. Magini, and T.H. de Freitas, *Laboratory gait analysis in patients with low back pain before and after a pilates intervention.* J Sport Rehabil, 2009. **18**(2): p. 269-82.

108. Gladwell, V., et al., *Does a program of Pilates improve chronic non-specific low back pain?* Journal of sport rehabilitation, 2006. **15**(4): p. 338.

109. Marshall, P.W., et al., *Pilates exercise or stationary cycling for chronic nonspecific low back pain: does it matter? a randomized controlled trial with 6-month follow-up.* Spine, 2013. **38**(15): p. E952-9.

110. Miyamoto, G.C., et al., *Efficacy of the addition of modified Pilates exercises to a minimal intervention in patients with chronic low back pain: a randomized controlled trial.* Phys Ther, 2013. **93**(3): p. 310-20.

111. Natour, J., et al., *Pilates improves pain, function and quality of life in patients with chronic low back pain: a randomized controlled trial.* Clin Rehabil, 2015. **29**(1): p. 59-68.

112. Quinn, K., S. Barry, and L. Barry, *Do patients with chronic low back pain benefit from attending Pilates classes after completing conventional physiotherapy treatment?* Physiotherapy Practice and Research, 2011. **32**(1): p. 5-12.

113. Rajpal, N., M. Arora, and V. Chauhan, *A Study on Efficacy of Pilates & Pilates & Mckenzie Exercises in Postural Low Back Pain-A Rehabilitative Protocol.* Physiotherapy and Occupational Therapy Journal, 2008. **1**(1): p. 33-56.

114. Rydeard, R., A. Leger, and D. Smith, *Pilates-based therapeutic exercise: effect on subjects with nonspecific chronic low back pain and functional disability: a randomized controlled trial.* Journal of orthopaedic & sports physical therapy, 2006. **36**(7): p. 472-484.

115. Wajswelner, H., B. Metcalf, and K. Bennell, *Clinical Pilates versus general exercise for chronic low back pain: randomized trial.* Med Sci Sports Exerc, 2012. **44**(7): p. 1197-205.

116. Adams, M.A., K. Burton, and N. Bogduk, *The biomechanics of back pain*. Vol. 55. 2006: Elsevier health sciences.

117. Bogduk, N., *Clinical anatomy of the lumbar spine and sacrum*. 2005: Elsevier Health Sciences.

118. Izzo, R., et al., *Biomechanics of the spine. Part I: spinal stability.* European journal of radiology, 2013. **82**(1): p. 118-126.

119. Panjabi, M.M., *The stabilizing system of the spine. Part I. Function, dysfunction, adaptation, and enhancement.* Journal of spinal disorders & techniques, 1992. **5**(4): p. 383-389.

120. Panjabi, M.M., *Clinical spinal instability and low back pain.* Journal of electromyography and kinesiology, 2003. **13**(4): p. 371-379.

121. Roberts, S., J. Menage, and J. Urban, *Biochemical and structural properties of the cartilage end-plate and its relation to the intervertebral disc.* Spine, 1989. **14**(2): p. 166-174.

122. Adams, M.A., *The mechanical properties of lumbar intervertebral joints with special reference to the causes of low back pain*. 1980, Polytechnic of Central London.

123. Adams, M. and W. Hutton, *The relevance of torsion to the mechanical derangement of the lumbar spine.* Spine, 1981. **6**(3): p. 241-248.

124. Adams, M.A. and P. Dolan, *Recent advances in lumbar spinal mechanics and their clinical significance.* Clinical Biomechanics, 1995. **10**(1): p. 3-19.

125. Ahmed, A.M., N.A. Duncan, and D.I. Burke, *The effect of facet geometry on the axial torque-rotation response of lumbar motion segments.* Spine, 1990. **15**(5): p. 391-401.

126. Cyron, B. and W. Hutton, *The behaviour of the lumbar intervertebral disc under repetitive forces.* International orthopaedics, 1981. **5**(3): p. 203-207.

127. Duncan, N.A. and A.M. Ahmed, *The Role of Axial Rotation in the Etiology of Unilateral Disc Prolapse: An Experimental and Finite-Element Analysis.* Spine, 1991. **16**(9): p. 1089-1098.

128. Gunzburg, R., W. Hutton, and R. Fraser, *Axial Rotation of the Lumbar Spine and the Effect of Flexion: An in Vitro and in Vivo Biomechanical Study.* Spine, 1991. **16**(1): p. 22-28.

129. Adams, M., P. Dolan, and W. Hutton, *Diurnal variations in the stresses on the lumbar spine.* Spine, 1987. **12**(2): p. 130-137.

130. Adams, M., W. Hutton, and J. Stott, *The resistance to flexion of the lumbar intervertebral joint.* Spine, 1980. **5**(3): p. 245-253.

131. Adams, M., et al., *The clinical biomechanics award paper 1993 posture and the compressive strength of the lumbar spine.* Clinical Biomechanics, 1994. **9**(1): p. 5-14.

132. Chazal, J., et al., *Biomechanical properties of spinal ligaments and a histological study of the supraspinal ligament in traction.* Journal of biomechanics, 1985. **18**(3): p. 167-176.

133. Panjabi, M.M., V.K. Goel, and K. Takata, *Physiologic Strains in the Lumbar Spinal Ligaments: An In VitroBiomechanical Study.* Spine, 1982. **7**(3): p. 192-203.

134. Schönström, N., et al., *Dynamic changes in the dimensions of the lumbar spinal canal: an experimental study in vitro.* Journal of orthopaedic research, 1989. **7**(1): p. 115-121.

135. Adams, M.A., T.P. Green, and P. Dolan, *The strength in anterior bending of lumbar intervertebral discs.* Spine, 1994. **19**(19): p. 2197-2203.

136. Brinckmann, P. and H. Grootenboer, *Change of Disc Height, Radial Disc Bulge, and Intradiscal Pressure From Discectomy An in Vitro Investigation on Human Lumbar Discs.* Spine, 1991. **16**(6): p. 641-646.

137. Setton, L.A., et al., *Compressive properties of the cartilaginous end-plate of the baboon lumbar spine.* Journal of Orthopaedic Research, 1993. **11**(2): p. 228-239.

138. Gray, H., *Grays Anatomy.* 1st. ed. 1858, London, UK.: The Promotional Reprint Company Limited.

139. McMinn, R., et al., *Wolfe Coloratlas A Colour Atlas of Human Anatomy.* 1993, London, UK.: Wolfe Publishing Ltd. .

140. Palastanga, N., D. Field, and R. Soames, *Anatomy & Human Movement - Structure & Function* 3rd. ed. 2000, Oxford, UK.: Butterworth Heinemann.

141. Aalto, T., *Preoperative predictors and postoperative outpatient rehabilitation of lumbar spinal stenosis - A two-year prospective follow-up.* 2013, University of Eastern Finland.

142. Kankaanpaa, M., et al., *Age, sex, and body mass index as determinants of back and hip extensor fatigue in the isometric Sorensen back endurance test.* Arch Phys Med Rehabil, 1998. **79**(9): p. 1069-75.

143. Dolan, P., M. Earley, and M. Adams, *Bending and compressive stresses acting on the lumbar spine during lifting activities.* Journal of biomechanics, 1994. **27**(10): p. 1237-1248.

144. Dolan, P., A. Mannion, and M. Adams, *Passive tissues help the back muscles to generate extensor moments during lifting.* Journal of Biomechanics, 1994. **27**(8): p. 1077-1085.

145. Frobin, W., et al., *Precision measurement of disc height, vertebral height and sagittal plane displacement from lateral radiographic views of the lumbar spine.* Clinical Biomechanics, 1997. **12**: p. S1-S63.

146. Macintosh, J.E., N. Bogduk, and M.J. Pearcy, *The effects of flexion on the geometry and actions of the lumbar erector spinae.* Spine, 1993. **18**(7): p. 884-893.

147. Kavcic, N., S. Grenier, and S.M. McGill, *Quantifying tissue loads and spine stability while performing commonly prescribed low back stabilization exercises.* Spine, 2004. **29**(20): p. 2319-2329.

148. Akuthota, V. and S.F. Nadler, *Core strengthening.* Archives of physical medicine and rehabilitation, 2004. **85**: p. 86-92.

149. Shumway-Cook, A. and M. Woollacott, *Motor Control - Translating Research into Clinical Practice*. 2007, Philadelphia, US.: Lippincott Wiliams & Wilkins.

150. Rosenbaum, D.A., *Human motor control*. 2009: Academic press.

151. Lundy Ekman, L., *Neuoscience, Fundamentals for Rehabilitation.* 2007, St. Louis, US.: Saunders Elsevier.

152. Solomonow, M., *Sensory-motor control of ligaments and associated neuromuscular disorders.* J Electromyogr Kinesiol, 2006. **16**(6): p. 549-67.

153. Apkarian, A.V., M.N. Baliki, and P.Y. Geha, *Towards a theory of chronic pain.* Prog Neurobiol, 2009. **87**(2): p. 81-97.

154. Baliki, M.N., et al., *Beyond feeling: chronic pain hurts the brain, disrupting the default-mode network dynamics.* J Neurosci, 2008. **28**(6): p. 1398-403.

155. Luomajoki, H., *Movement control impairment as a sub-group of non-specific low back pain. Evaluation of movement control test battery as a practical tool in the diagnosis of movement control impairment and treatment of this dysfunction*. 2010, University of Eastern Finland.

156. Flor, H., *Remapping somatosensory cortex after injury.* Adv Neurol, 2003. **93**: p. 195-204.

157. Flor, H., et al., *Extensive reorganization of primary somatosensory cortex in chronic back pain patients.* Neurosci Lett, 1997. **224**(1): p. 5-8.

158. Flor, H., et al., *Effect of sensory discrimination training on cortical reorganisation and phantom limb pain.* The Lancet, 2001. **357**(9270): p. 1763-1764.

159. Moseley, G.L., *I can't find it! Distorted body image and tactile dysfunction in patients with chronic back pain.* Pain, 2008. **140**(1): p. 239-43.

160. Moseley, G.L., *Pain, brain imaging and physiotherapy--opportunity is knocking.* Man Ther, 2008. **13**(6): p. 475-7.

161. Moseley, G.L., A. Gallace, and C. Spence, *Is mirror therapy all it is cracked up to be? Current evidence and future directions.* Pain, 2008. **138**(1): p. 7-10.

162. Pleger, B., et al., *Patterns of cortical reorganization parallel impaired tactile discrimination and pain intensity in complex regional pain syndrome.* NeuroImage, 2006. **32**(2): p. 503-510.

163. Wand, B.M., et al., *Cortical changes in chronic low back pain: current state of the art and implications for clinical practice.* Man Ther, 2011. **16**(1): p. 15-20.

164. Shumway-Cook, A., D. Anson, and S. Haller, *Postural sway biofeedback: its effect on reestablishing stance stability in hemiplegic patients.* Arch Phys Med Rehabil, 1988. **69**(6): p. 395-400.

165. Rast, F.M., et al., *Reproducibility of a new signal processing technique to assess joint sway during standing.* Journal of Biomechanics, 2017. **51**: p. 133-136.

166. Gribble, P.A. and J. Hertel, *Effect of lower-extremity muscle fatigue on postural control.* Arch Phys Med Rehabil, 2004. **85**(4): p. 589-92.

167. Davidson, B.S., M.L. Madigan, and M.A. Nussbaum, *Effects of lumbar extensor fatigue and fatigue rate on postural sway.* Eur J Appl Physiol, 2004. **93**(1-2): p. 183-9.

168. Loram, I., *Postural Control and Sensorimotor Integration.*, in *Grieve's modern musculoskeletal physiotherapy.* , G.A. Jull, et al., Editors. 2015, Elsevier Ltd. : Edinburgh, UK.

169. Haber, S.N., J.L. Fudge, and N.R. McFarland, *Striatonigrostriatal Pathways in Primates Form an Ascending Spiral from the Shell to the Dorsolateral Striatum.* The Journal of Neuroscience, 2000. **20**(6): p. 2369-2382.

170. van de Kamp, C., et al., *Interfacing sensory input with motor output: does the control architecture converge to a serial process along a single channel?* Frontiers in Computational Neuroscience, 2013. **7**: p. 55.

171. Pruszynski, J.A. and S.H. Scott, *Optimal feedback control and the long-latency stretch response.* Experimental Brain Research, 2012. **218**(3): p. 341-359.

172. Sung, P.S. and M.J. Maxwell, *Kinematic chain reactions on trunk and dynamic postural steadiness in subjects with recurrent low back pain.* Journal of Biomechanics, 2017. **59**: p. 109-115.

173. Tsao, H., M.P. Galea, and P.W. Hodges, *Reorganization of the motor cortex is associated with postural control deficits in recurrent low back pain.* Brain, 2008. **131**(8): p. 2161-2171.

174. Claeys, K., et al., *Decreased variability in postural control strategies in young people with non-specific low back pain is associated with altered proprioceptive reweighting.* European Journal of Applied Physiology, 2011. **111**(1): p. 115-123.

175. Tsao, H., L.A. Danneels, and P.W. Hodges, *ISSLS prize winner: Smudging the motor brain in young adults with recurrent low back pain.* Spine (Phila Pa 1976), 2011. **36**(21): p. 1721-7.

176. van Dieen, J.H., J. Cholewicki, and A. Radebold, *Trunk muscle recruitment patterns in patients with low back pain enhance the stability of the lumbar spine.* Spine (Phila Pa 1976), 2003. **28**(8): p. 834-41.

177. Sung, P.S., B. Yoon, and D.C. Lee, *Lumbar Spine Stability for Subjects With and Without Low Back Pain During One-Leg Standing Test.* Spine, 2010. **35**(16): p. E753-E760.

178. Brumagne, S., P. Cordo, and S. Verschueren, *Proprioceptive weighting changes in persons with low back pain and elderly persons during upright standing.* Neuroscience Letters, 2004. **366**(1): p. 63-66.

179. Hodges, P., et al., *Changes in the mechanical properties of the trunk in low back pain may be associated with recurrence.* J Biomech, 2009. **42**(1): p. 61-6.

180. Solomonow, M., *Time dependent spine stability: the wise old man and the six blind elephants.* Clin Biomech, 2011. **26**(3): p. 219-28.

181. Solomonow, M., *Neuromuscular manifestations of viscoelastic tissue degradation following high and low risk repetitive lumbar flexion.* J Electromyogr Kinesiol, 2012. **22**(2): p. 155-75.

182. Holtermann, A., et al., *Patient handling and risk for developing persistent low-back pain among female healthcare workers.* Scand J Work Environ Health, 2013. **39**(2): p. 164-9.

183. Roffey, D.M., et al., *Causal assessment of workplace manual handling or assisting patients and low back pain: results of a systematic review.* Spine J, 2010. **10**(7): p. 639-51.

184. Seidler, A., et al., *Physical workload and accelerated occurrence of lumbar spine diseases: risk and rate advancement periods in a German multicenter case-control study.* Scand J Work Environ Health, 2011. **37**(1): p. 30-6.

185. Smedley, J., et al., *Manual handling activities and risk of low back pain in nurses.* Occup Environ Med, 1995. **52**(3): p. 160-3.

186. Yassi, A. and K. Lockhart, *Work-relatedness of low back pain in nursing personnel: a systematic review.* Int J Occup Environ Health, 2013. **19**(3): p. 223-44.

187. Gooyers, C.E., et al., *The impact of posture and prolonged cyclic compressive loading on vertebral joint mechanics.* Spine, 2012. **37**(17): p. E1023-9.

188. Marras, W.S., et al., *Cumulative spine loading and clinically meaningful declines in low-back function.* Hum Factors, 2014. **56**(1): p. 29-43.

189. Hodder, J.N., M.W. Holmes, and P.J. Keir, *Continuous assessment of work activities and posture in long-term care nurses.* Ergonomics, 2010. **53**(9): p. 1097-107.

190. Holmes, M.W., J.N. Hodder, and P.J. Keir, *Continuous assessment of low back loads in long-term care nurses.* Ergonomics, 2010. **53**(9): p. 1108-16.

191. Marras, W., *Occupational low back disorder causation and control.* Ergonomics, 2000. **43**(7): p. 880-902.

192. Yang, G., W.S. Marras, and T.M. Best, *The biochemical response to biomechanical tissue loading on the low back during physical work exposure.* Clin Biomech, 2011. **26**(5): p. 431-7.

193. Solomonow, M., et al., *Acute repetitive lumbar syndrome: a multi-component insight into the disorder.* J Bodyw Mov Ther, 2012. **16**(2): p. 134-47.

194. Ben-Masaud, A., et al., *Motor control of lumbar instability following exposure to various cyclic load magnitudes.* European Spine Journal, 2009. **18**(7): p. 1022-1034.

195. D'Ambrosia, P., et al., *Increase pro-inflammatory cytokines in lumbar ligaments following low and high magnitudes cyclic loading.* Eur Spine J, 2010. **19**: p. 1330-1339.

196. Eversull, B.E., et al., *Neuromuscular neutral zones sensitivity to lumbar displacement rate.* Clinical Biomechanics, 2001. **16**(2): p. 102-113.

197. Hoops, H., et al., *Short rest between cyclic flexion periods is a risk factor for a lumbar disorder.* Clinical Biomechanics, 2007. **22**(7): p. 745-757.

198. King, K., et al., *High magnitude cyclic load triggers inflammatory response in lumbar ligaments.* Clinical Biomechanics, 2009. **24**(10): p. 792-798.

199. Le, P., et al., *Cyclic load magnitude is a risk factor for a cumulative lower back disorder.* Journal of Occupational and Environmental Medicine, 2007. **49**(4): p. 375-387.

200. Lu, D., et al., *Frequency-dependent changes in neuromuscular responses to cyclic lumbar flexion.* Journal of biomechanics, 2004. **37**(6): p. 845-855.

201. Navar, D., et al., *High-repetition cyclic loading is a risk factor for a lumbar disorder.* Muscle & nerve, 2006. **34**(5): p. 614-622.

202. Pinski, S.E., et al., *High-frequency loading of lumbar ligaments increases proinflammatory cytokines expression in a feline model of repetitive musculoskeletal disorder.* The Spine Journal, 2010. **10**(12): p. 1078-1085.

203. Solomonow, D., et al., *Neuromuscular neutral zones response to cyclic lumbar flexion.* Journal of biomechanics, 2008. **41**(13): p. 2821-2828.

204. Solomonow, M., et al., *Neuromuscular neutral zones associated with viscoelastic hysteresis during cyclic lumbar flexion.* Spine, 2001. **26**(14): p. E314-E324.

205. Cholewicki, J., S.M. McGill, and R.W. Norman, *Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: towards development of a hybrid approach.* Journal of biomechanics, 1995. **28**(3): p. 321-331.

206. Granata, K.P. and W. Marras, *The influence of trunk muscle coactivity on dynamic spinal loads.* Spine, 1995. **20**(8): p. 913-919.

207. Kingma, I., et al., *Lumbar loading during lifting: a comparative study of three measurement techniques.* Journal of Electromyography and Kinesiology, 2001. **11**(5): p. 337-345.

208. Lavender, S.A., et al., *Coactivation of the trunk muscles during asymmetric loading of the torso.* Human Factors: The Journal of the Human Factors and Ergonomics Society, 1992. **34**(2): p. 239-247.

209. McGill, S.M., *Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics.* Journal of Orthopaedic Research, 1991. **9**(1): p. 91-103.

210. Stairmand, J., S. Holm, and J. Urban, *Factors Influencing Oxygen Concentration Gradients in the Intervertebral Disc: A Theoretical Analysis.* Spine, 1991. **16**(4): p. 444-449.

211. Cassisi, J.E., et al., *Trunk strength and lumbar paraspinal muscle activity during isometric exercise in chronic low-back pain patients and controls.* Spine, 1993. **18**(2): p. 245-251.

212. Basmajian, J.V., *Acute back pain and spasm. A controlled multicenter trial of combined analgesic and antispasm agents.* Spine, 1989. **14**(4): p. 438-439.

213. Solomonow, M., et al., *Biomechanics of increased exposure to lumbar injury caused by cyclic loading: Part 1. Loss of reflexive muscular stabilization.* Spine, 1999. **24**(23): p. 2426-2434.

214. Watson, P., et al., *Surface electromyography in the identification of chronic low back pain patients: the development of the flexion relaxation ratio.* Clinical Biomechanics, 1997. **12**(3): p. 165-171.

215. Stubbs, M., et al., *Ligamento-muscular protective reflex in the lumbar spine of the feline.* Journal of Electromyography and Kinesiology, 1998. **8**(4): p. 197-204.

216. Solomonow, M., et al., *Biexponential recovery model of lumbar viscoelastic laxity and reflexive muscular activity after prolonged cyclic loading.* Clinical Biomechanics, 2000. **15**(3): p. 167-175.

217. Hides, J., et al., *Evidence of lumbar multifidus muscle wasting ipsilateral to symptoms in patients with acute/subacute low back pain.* Spine, 1994. **19**(2): p. 165-172.

218. Stokes, M. and A. Young, *The contribution of reflex inhibition to arthrogenous muscle weakness.* Clinical Science, 1984. **67**(1): p. 7-14.

219. Laird, R.A., et al., *Comparing lumbo-pelvic kinematics in people with and without back pain: a systematic review and meta-analysis.* Bmc Musculoskeletal Disorders, 2014. **15**(1): p. 229.

220. Rast, F.M., et al., *Between-day reliability of three-dimensional motion analysis of the trunk: A comparison of marker based protocols.* Journal of Biomechanics, 2016. **49**(5): p. 807-811.

221. Roetenberg, D., C.T. Baten, and P.H. Veltink, *Estimating body segment orientation by applying inertial and magnetic sensing near ferromagnetic materials.* IEEE Trans Neural Syst Rehabil Eng, 2007. **15**(3): p. 469-71.

222. Slaboda, J.C., et al., *The use of splines to calculate jerk for a lifting task involving chronic lower back pain patients.* IEEE Trans Neural Syst Rehabil Eng, 2005. **13**(3): p. 406-14.

223. Dideriksen, J.L., et al., *Deterministic accessory spinal movement in functional tasks characterizes individuals with low back pain.* Clin Neurophysiol, 2014. **125**(8): p. 1663-8.

224. Madeleine, P. and T.M. Madsen, *Changes in the amount and structure of motor variability during a deboning process are associated with work experience and neck-shoulder discomfort.* Appl Ergon, 2009. **40**(5): p. 887-94.

225. Rausch Osthoff, A.K., et al., *Measuring Lumbar Reposition Accuracy in Patients With Unspecific Low Back Pain: Systematic Review and Meta-analysis.* Spine, 2015. **40**(2): p. E97-E111.

226. Association, A.P., A.E.R. Association, and N.C.o.M.i. Education, *Technical recommendations for psychological tests and diagnostic techniques*. Vol. 51. 1954: American Psychological Association.

227. De Vet, H.C., et al., *Measurement in medicine: a practical guide*. 2011: Cambridge University Press.

228. de Vet, H.C., et al., *When to use agreement versus reliability measures.* J Clin Epidemiol, 2006. **59**(10): p. 1033-9.

229. Corriveau, H., et al., *Intrasession reliability of the "center of pressure minus center of mass" variable of postural control in the healthy elderly.* Archives of Physical Medicine and Rehabilitation, 2000. **81**(1): p. 45-48.

230. Ernst, M.J., et al., *Determination of thoracic and lumbar spinal processes by their percentage position between C7 and the PSIS level.* BMC Res Notes, 2013. **6**: p. 58.

231. Santos, B.R., et al., *Reliability of centre of pressure summary measures of postural steadiness in healthy young adults.* Gait Posture, 2008. **27**(3): p. 408-15.

232. Hocoma, A., *Valedo®Motion User Manual.*, ed. H. AG. Vol. 2nd edition. 2011, Volketswil.

233. Schelldorfer, S., et al., *Low back pain and postural control, effects of task difficulty on centre of pressure and spinal kinematics.* Gait Posture, 2015. **41**(1): p. 112-8.

234. Madgwick, S., R. Vaidyanathan, and A. Harrison, *An Efficient Orientation Filter for IMU and MARG Sensor Arrays*. 2010, Department of Mechanical Engineering, University of Bristol.

235. Crawford, N.R., G.T. Yamaguchi, and C.A. Dickman, *A new technique for determining 3-D joint angles: the tilt/twist method.* Clin Biomech, 1999. **14**(3): p. 153-65.

236. Bauer, C.M., et al., *Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement.* J Electromyogr Kinesiol, 2015. **25**(5): p. 782-90.

237. Bauer, C.M., et al., *The effect of muscle fatigue and low back pain on lumbar movement variability and complexity.* J Electromyogr Kinesiol, 2017. **33**: p. 94-102.

238. Bauer, C.M., et al., *Pain intensity attenuates movement control of the lumbar spine in low back pain.* J Electromyogr Kinesiol, 2015. **25**(6): p. 919-27.

239. Marwan, N. and J. Kurths, *Nonlinear analysis of bivariate data with cross recurrence plots.* Physics Letters A, 2002. **302**(5–6): p. 299-307.

240. Lee, K. *Sample Entropy*. File Exchange 2012 [cited 2012 11.07.2016]; Available from: https://www.mathworks.com/matlabcentral/fileexchange/35784-sample-entropy.

241. Hogan, N. and D. Sternad, *Sensitivity of smoothness measures to movement duration, amplitude, and arrests.* J Mot Behav, 2009. **41**(6): p. 529-34.

242. Webber, C.L., Jr. and J.P. Zbilut, *Dynamical assessment of physiological systems and states using recurrence plot strategies.* J Appl Physiol, 1994. **76**(2): p. 965-73.

243. Webber, C.L., Jr. and N. Marwan, *Recurrence Quantification Analysis*. Vol. 1. 2015: Springer.

244. Richman, J.S. and J.R. Moorman, *Physiological time-series analysis using approximate entropy and sample entropy.* Am J Physiol Heart Circ Physiol, 2000. **278**(6): p. H2039-49.

245. Biering-Sorensen, F., *Physical measurements as risk indicators for low-back trouble over a one-year period.* Spine, 1984. **9**(2): p. 106-19.

246. Suni, J., et al., *Neuromuscular exercises and back counseling for female nursing personnel with non-chronic nonspecific low back pain: Study protcol of a ranomized controlled trial NCT4165698*. BMJ Open Sport & Exercise Medicine, 2016.

247. Consmuller, T., et al., *Velocity of lordosis angle during spinal flexion and extension.* PLoS One, 2012. **7**(11): p. e50135.

248. Galati-Petrecca, M., *Swiss Health Survey 2007, First findings*, F.S.O. Swiss Confederation, Editor. 2008, Federal Statistical Office (FSO): Section of Population Health – Swiss Health Survey.

249. Brennan, R.L., *Generalizability Theory*. Statistics for Social Science and Public Policy, ed. S.E. Fienberg, D. Lievesley, and J. Rolph. 2001, Berlin Heidelberg New York: Springer-Verlag.

250. Carter, R.E., J. Lubinsky, and E. Domholdt, *Rehabilitation Research: Principles and Applications*. 4th ed. 2005, St. Louis, MO: Elsevier Saunders.

251. Hopkins, W.G., *Measures of reliability in sports medicine and science.* Sports Med, 2000. **30**(1): p. 1-15.

252. Suni, J., M. Rinne, and J. Ruiz, *Retest Repeatability of Motor and Musculoskeletal Fitness Tests for Public Health Monitoring of Adult Populations.* Journal of Novel Physiotherapies, 2014. **4**(1).

253. Plummer, M. *JAGS: A program for analysis of Bayesian graphical models using Gibbs sampling.* in *Proceedings of the 3rd International Workshop on Distributed Statistical Computing (DSC 2003)*. 2003.

254. Little, R. and D. Rubin, *Statistical Analysis with Missing Data*. 2nd ed. WILEY SERIES IN PROBABILITY AND STATISTICS, ed. I. JOHN WILEY & SONS. 2002, Hoboken, New Jersey, USA.: JOHN WILEY & SONS, INC.

255. Diggle, P., et al., *Analysis of longitudinal data. 2002*. Oxford Statistical Science Series, 2002.

256. Bauer, C., et al., *Efficacy of six months neuromuscular exercise on lumbar movement variability – A randomized controlled trial.* in preparation.

257. Bauer, C.M., et al., *Reliability of lumbar movement dysfunction tests for chronic low back pain patients.* Man Ther, 2016. **24**: p. 81-4.

258. Luomajoki, H., et al., *Movement control tests of the low back; evaluation of the difference between patients with low back pain and healthy controls.* BMC Musculoskeletal Disorders, 2008. **9**(1): p. 170.

259. Lipsitz, L.A., *Dynamics of stability: the physiologic basis of functional health and frailty.* J Gerontol A Biol Sci Med Sci, 2002. **57**(3): p. B115-25.

260. Lipsitz, L.A. and A.L. Goldberger, *Loss of 'complexity' and aging. Potential applications of fractals and chaos theory to senescence.* Jama, 1992. **267**(13): p. 1806-9.

261. Stergiou, N. and L.M. Decker, *Human movement variability, nonlinear dynamics, and pathology: is there a connection?* Hum Mov Sci, 2011. **30**(5): p. 869-88.

262. Madeleine, P., *On functional motor adaptations: from the quantification of motor strategies to the prevention of musculoskeletal disorders in the neck-shoulder region.* Acta Physiol, 2010. **199 Suppl 679**: p. 1-46.

263. Ha, T.H., et al., *Measurement of lumbar spine range of movement and coupled motion using inertial sensors - a protocol validity study.* Man Ther, 2013. **18**(1): p. 87-91.

264. Dunne, L.E., et al. *Design and Evaluation of a Wearable Optical Sensor for Monitoring Seated Spinal Posture.* in *Wearable Computers, 2006 10th IEEE International Symposium on*. 2006.

265. Wong, W.Y. and M.S. Wong, *Trunk posture monitoring with inertial sensors.* Eur Spine J, 2008. **17**(5): p. 743-53.

266. Luomajoki, H., et al., *Reliability of movement control tests in the lumbar spine.* BMC Musculoskeletal Disorders, 2007. **8**(1): p. 90.

267. Lehtola, V., et al., *Sub-classification based specific movement control exercises are superior to general exercise in sub-acute low back pain when both are combined with manual therapy: A randomized controlled trial.* BMC Musculoskelet Disord, 2016. **17**: p. 135.

268. Saner, J., et al., *A tailored exercise program versus general exercise for a subgroup of patients with low back pain and movement control impairment: A randomised controlled trial with one-year follow-up.* Man Ther, 2015.

269. Lamoth, C.C., et al., *Effects of chronic low back pain on trunk coordination and back muscle activity during walking: changes in motor control.* European Spine Journal, 2006. **15**(1): p. 23-40.

270. Silfies, S.P., et al., *Trunk control during standing reach: A dynamical system analysis of movement strategies in patients with mechanical low back pain.* Gait & Posture, 2009. **29**(3): p. 370-376.

271. Falla, D., et al., *Reduced task-induced variations in the distribution of activity across back muscle regions in individuals with low back pain.* Pain, 2014. **155**(5): p. 944-53.

272. Arzi, H., et al., *Movement control in patients with shoulder instability: a comparison between patients after open surgery and nonoperated patients.* Journal of Shoulder and Elbow Surgery, 2014. **23**(7): p. 982-992.

273. Uri, O., et al., *Upper limb kinematics after arthroscopic and open shoulder stabilization.* J Shoulder Elbow Surg, 2015. **24**(3): p. 399-406.

274. Price, D.D., et al., *A comparison of pain measurement characteristics of mechanical visual analogue and simple numerical rating scales.* Pain, 1994. **56**(2): p. 217-26.

275. Cignetti, F., F. Schena, and A. Rouard, *Effects of fatigue on inter-cycle variability in cross-country skiing.* J Biomech, 2009. **42**(10): p. 1452-9.

276. Cote, J.N., et al., *Effects of fatigue on intermuscular coordination during repetitive hammering.* Motor Control, 2008. **12**(2): p. 79-92.

277. Cote, J.N., et al., *Differences in multi-joint kinematic patterns of repetitive hammering in healthy, fatigued and shoulder-injured individuals.* Clin Biomech, 2005. **20**(6): p. 581-90.

278. Forestier, N. and V. Nougier, *The effects of muscular fatigue on the coordination of a multijoint movement in human.* Neurosci Lett, 1998. **252**(3): p. 187-90.

279. Fuller, J.R., J. Fung, and J.N. Cote, *Time-dependent adaptations to posture and movement characteristics during the development of repetitive reaching induced fatigue.* Exp Brain Res, 2011. **211**(1): p. 133-43.

280. Huffenus, A.F., D. Amarantini, and N. Forestier, *Effects of distal and proximal arm muscles fatigue on multi-joint movement organization.* Exp Brain Res, 2006. **170**(4): p. 438-47.

281. Selen, L.P., P.J. Beek, and J.H. van Dieen, *Fatigue-induced changes of impedance and performance in target tracking.* Exp Brain Res, 2007. **181**(1): p. 99-108.

282. Srinivasan, D. and S.E. Mathiassen, *Motor variability in occupational health and performance.* Clinical biomechanics, 2012. **27**(10): p. 979-993.

283. Falla, D., L. Arendt-Nielsen, and D. Farina, *The pain-induced change in relative activation of upper trapezius muscle regions is independent of the site of noxious stimulation.* Clin Neurophysiol, 2009. **120**(1): p. 150-7.

284. Farina, D., et al., *Low-frequency oscillations of the neural drive to the muscle are increased with experimental muscle pain.* J Neurophysiol, 2012. **107**(3): p. 958-65.

285. Muceli, S., D. Falla, and D. Farina, *Reorganization of muscle synergies during multidirectional reaching in the horizontal plane with experimental muscle pain.* J Neurophysiol, 2014. **111**(8): p. 1615-30.

286. Yavuz, U.S., et al., *Experimental muscle pain increases variability of neural drive to muscle and decreases motor unit coherence in tremor frequency band.* J Neurophysiol, 2015. **114**(2): p. 1041-7.

287. Felici, F., et al., *Linear and non-linear analysis of surface electromyograms in weightlifters.* European Journal of Applied Physiology, 2001. **84**(4): p. 337-342.

288. Van Den Hoorn, W., et al., *Mechanical coupling between transverse plane pelvis and thorax rotations during gait is higher in people with low back pain.* Journal of Biomechanics, 2012. **45**(2): p. 342-347.

289. Gates, D.H. and J.B. Dingwell, *The effects of neuromuscular fatigue on task performance during repetitive goal-directed movements.* Exp Brain Res, 2008. **187**(4): p. 573-85.

290. van Dieen, J.H., et al., *Low-level activity of the trunk extensor muscles causes electromyographic manifestations of fatigue in absence of decreased oxygenation.* J Electromyogr Kinesiol, 2009. **19**(3): p. 398-406.

291. Lomond, K.V. and J.N. Cote, *Movement timing and reach to reach variability during a repetitive reaching task in persons with chronic neck/shoulder pain and healthy subjects.* Exp Brain Res, 2010. **206**(3): p. 271-82.

292. Costa, M., A.L. Goldberger, and C.K. Peng, *Multiscale entropy analysis of biological signals.* Phys Rev E Stat Nonlin Soft Matter Phys, 2005. **71**(2 Pt 1): p. 021906.

293. Samani, A., et al., *Active biofeedback changes the spatial distribution of upper trapezius muscle activity during computer work.* Eur J Appl Physiol, 2010. **110**(2): p. 415-23.

294. Granata, K.P., W.S. Marras, and K.G. Davis, *Variation in spinal load and trunk dynamics during repeated lifting exertions.* Clin Biomech, 1999. **14**(6): p. 367-75.

295. Hasson, C.J., et al., *Influence of embedding parameters and noise in center of pressure recurrence quantification analysis.* Gait & posture, 2008. **27**(3): p. 416-422.

296. Ferreira-Valente, M.A., J.L. Pais-Ribeiro, and M.P. Jensen, *Validity of four pain intensity rating scales.* Pain, 2011. **152**(10): p. 2399-404.

297. Sackett, D.L. and R.B. Haynes, *The architecture of diagnostic research.* Bmj, 2002. **324**(7336): p. 539-541.

298. Bolink, S.A.A.N., et al., *Validity of an inertial measurement unit to assess pelvic orientation angles during gait, sit–stand transfers and step-up transfers: Comparison with an optoelectronic motion capture system*.* Medical Engineering & Physics, 2016. **38**(3): p. 225-231.

299. Matheve, T., S. Brumagne, and A.A. Timmermans, *The effectiveness of technology-supported exercise therapy for low back pain: A systematic review.* American journal of physical medicine & rehabilitation, 2017. **96**(5): p. 347-356.

300. *Bundesärztekammer (BÄK), Kassenärztliche Bundesvereinigung (KBV), Arbeitsgemeinschaft der Wissenschaftlichen Medizinischen Fachgesellschaften (AWMF) (2010, zuletzt verändert: 2013.) Nationale VersorgungsLeitlinie Kreuzschmerz – Langfassung. Version 4. doi:10.6101/AZQ/000149.*

301. Niederer, D., et al., *Medicine in spine exercise (MiSpEx) for nonspecific low back pain patients: study protocol for a multicentre, single-blind randomized controlled trial.* Trials, 2016. **17**(1): p. 507.

302. Moseley, G.L., *Innovative treatments for back pain.* Pain, 2017. **158**: p. S2-S10.

303. Heinrich, M., S. Steiner, and C. Bauer, *The influence of visual feedback on chronic non specific back pain. A systematic review.* Physiotherapy Research and Practice, under review.

304. Friedrich, M., et al., *Combined exercise and motivation program: effect on the compliance and level of disability of patients with chronic low back pain: a randomized controlled trial.* Archives of physical medicine and rehabilitation, 1998. **79**(5): p. 475-487.

305. Härkäpää, K., et al., *Health locus of control beliefs and psychological distress as predictors for treatment outcome in low-back pain patients: results of a 3-month follow-up of a controlled intervention study.* Pain, 1991. **46**(1): p. 35-41.

306. Mannion, A.F., et al., *Spinal segmental stabilisation exercises for chronic low back pain: programme adherence and its influence on clinical outcome.* European spine journal, 2009. **18**(12): p. 1881-1891.

307. Marios, T., N.A. Smart, and S. Dalton, *The effect of tele-monitoring on exercise training adherence, functional capacity, quality of life and glycemic control in patients with type II diabetes.* Journal of sports science & medicine, 2012. **11**(1): p. 51.

308. Watson, A., et al., *An internet-based virtual coach to promote physical activity adherence in overweight adults: randomized controlled trial.* Journal of medical Internet research, 2012. **14**(1): p. e1.

309. Hügli, A.S., et al., *Adherence to home exercises in non-specific low back pain. A randomised controlled pilot trial.* Journal of bodywork and movement therapies, 2015. **19**(1): p. 177-185.

310. B.-N. Sanders, E., *From user-centered to participatory design approaches*, in *Design and the social sciences: Making connections*. 2002, CRC Press. p. 1-8.

311. Ortiz, J., et al., *XoSoft-A Vision for a Soft Modular Lower Limb Exoskeleton*, in *Wearable Robotics: Challenges and Trends*. 2017, Springer. p. 83-88.

312. Kennel, M.B., R. Brown, and H.D. Abarbanel, *Determining embedding dimension for phase-space reconstruction using a geometrical construction.* Physical review A, 1992. **45**(6): p. 3403.

313. Rissanen, S.M., et al., *Analysis of dynamic EMG and acceleration measurements in Parkinson's disease.* Conf Proc IEEE Eng Med Biol Soc, 2008. **2008**: p. 5053-6.

314. Edwards, J.R. and R.P. Bagozzi, *On the nature and direction of relationships between constructs and measures.* Psychological methods, 2000. **5**(2): p. 155.

315. Webber, C.L., Jr., C. Ioana, and N. Marwan. *Recurrence Plots and Their Quantifications: Expanding Horizoms*. in *6th International Symposium on Recurrence Plots*. 2015. Grenoble, France: Springer.

316. Guhathakurta, K., B. Bhattacharya, and A.R. Chowdhury, *Using recurrence plot analysis to distinguish between endogenous and exogenous stock market crashes.* Physica A: Statistical Mechanics and its Applications, 2010. **389**(9): p. 1874-1882.

317. Monk, A. and A. Compton, *Recurrence phenomena in cosmic-ray intensity.* Reviews of Modern Physics, 1939. **11**(3-4): p. 173.

318. Eckmann, J.-P., S.O. Kamphorst, and D. Ruelle, *Recurrence plots of dynamical systems.* EPL (Europhysics Letters), 1987. **4**(9): p. 973.

319. Bigdeli, N., M. Jafarzadeh, and K. Afshar. *Characterization of Iran Stock Market Indices Using Recurrence Plots*. in *2011 International Conference on Management and Service Science*. 2011.

320. Aßmann, B., et al., *Hierarchical organization of a reference system in newborn spontaneous movements.* Infant Behavior and Development, 2007. **30**(4): p. 568-586.

10 APPENDIX

10.1 Detailed Overview of Movement Tests

Table 10. Overview of movement tests. 4pk = four-point kneeling; bpm = beats per minute; CE = constant error; DET = determinism; LS = lumbar spine; MCI = movement control impairment; RE = reposition error; REC = recurrence; RM = repetitive movement; rmsj = root mean squared jerk; ROM = range of motion; SaEn = sample entropy; SIT = sitting; TS = thoracic spine

Test	Starting Position	Movement	Body Segment of Interest	Variables (units)	Instructions	Remarks
ROM Tests						Participants performed the ROM tests at their own preferred speed. No target speeds or ranges for the moving segment were given.
ROM Flexion	Standing upright	Maximal flexion of the LS	TS, LS, hip	ROM TS, LS, Hip (°)	Stand upright and flex their lumbar spine as far as possible without flexing the hips.	
ROM Extension	Standing upright	Maximal extension of the LS	TS, LS, hip	ROM TS, LS, Hip (°)	Stand upright and extend their lumbar spine as far as possible without extending the hips.	
ROM Lateral Bending Right	Standing upright	Maximal lateral flexion of the LS	TS, LS, hip	ROM TS, LS, Hip (°)	Stand upright and lateral flex their lumbar spine as far as possible to the right without abducting the hips.	
ROM Lateral Bending Left	Standing upright	Maximal lateral flexion of the LS	TS, LS, hip	ROM TS, LS, Hip (°)	Stand upright and lateral flex their lumbar spine as far as possible to the left without abducting the hips.	

MCI Tests						The participants performed all MCI tests at their own preferred speed. No target range for the moving segment, or target speed was given. They were asked to terminate the movement before they perceived movement of their lumbar spine. During pelvic tilt they were asked to terminate movement before they perceived movement of their thoracic spine.
Pelvic Tilt	Standing upright	Anterior pelvic tilt without moving the trunk or knees	TS, LS	Ratio ROM TS/ROM LS rmsj TS (°/s³) rmsj LS (°/s³)	Stand upright and then tilt the pelvis anteriorly as far as possible whilst keeping the thoracic spine stable.	
Sitting Knee Extension	Sitting upright; hips at 90°	Knee extension without moving the LS	LS	ROM LS (°) rmsj LS (°/s³)	Stabilize the lumbar spine whilst extending the right knee.	
Waiter's Bow	Standing upright	Hip flexion without moving the LS	LS, hip	Ratio ROM LS/ROM Hip rmsj LS (°/s³) rmsj Hip (°/s³)	Stand upright and then flex the hips as far as possible whilst keeping the lumbar spine stable.	
Rocking Backwards	4pk	Hip flexion and shoulder extension without moving the LS	LS, hip	Ratio ROM LS/ROM Hip rmsj LS (°/s³) rmsj Hip (°/s³)	Stabilize the lumbar spine whilst flexing the hips and extending the shoulders.	
Rocking Forwards	4pk	Hip extension and shoulder flexion without moving the LS	LS, hip	Ratio ROM LS/ROM Hip rmsj LS (°/s³) rmsj Hip (°/s³)	Stabilize the lumbar spine whilst extending the hips and flexing the shoulders.	
Prone Knee Bend	Lying prone	Knee flexion without moving the LS	LS	ROM LS (°) rmsj LS (°/s³)	Stabilize the lumbar spine whilst flexing the right knee.	
RM Tests						The participants performed the RM tests at a predefined speed (by a metronome)
Picking Up a Box I & II	Standing upright	Lifting a box (10% body weight)	LS	REC, DET, and SaEn of angular displacement, velocity, and acceleration	During each cycle, participants were asked to pick up the box from the ground and put it back down within two beats, respectively. They were guided with a metronome. The box was placed at a standardized distance in front of them.	Study II: 5 cycles of 4 sec duration; Metronome 60bpm Study III: 10 cycles of 4 sec duration; Metronome 60bpm Study V: 5 cycles of 4.8 sec duration; Metronome 50bpm

Repeated Flexion and Extension	Sitting upright with hips at 60°	Repeated flexion and extension of the trunk	LS	REC, DET and SaEn of angular displacement, velocity and acceleration	During each cycle, participants were asked to flex and extend their lumbar spine and hip as far as possible whilst adhering to the timed window. Guided with a metronome, they were instructed to flex and extend within two beats, respectively. They were fixed at their thighs with two belts to prevent unintended movements of pelvis and thighs.	Study II: 5 cycles of 3 sec duration; Metronome 80bpm Study IV: 20 cycles of 3 sec duration; Metronome 80bpm
RE Tests						Participants performed the RE tests blindfolded, at their own preferred speed. The audio signal sounded at 50% of their maximal ROM, measured prior to RE tests.
Trunk Flexion	Sitting upright hips at 60°	Flexion of the trunk and reproducing the starting position	LS	CE LS (°)	"Flex the trunk until you hear an audio signal. After that return to the starting position and this position as accurately as possible."	
Posterior Pelvic Tilt	4pk	Extension of the LS and reproducing the starting position	LS	CE LS (°)	"Extend the lumbar spine until you hear an audio signal. After that return to the starting position and reproduce this position as accurately as possible."	

10.2 Recurrence Quantification Analysis and Sample Entropy

Recurrence quantification is an advanced technique of nonlinear data analysis based on a recurrence plot ($RP_{i,j}$) [239]. An $RP_{i,j}$ is a graph of a square matrix in which the matrix elements correspond to the times at which a state of a dynamical system recurs (columns and rows correspond then to a certain pair of times) [239, 315]. Thus, an $RP_{i,j}$ illustrates the times when a phase space trajectory of a dynamical system, such as lumbar movement data, visits a neighbouring area in a phase space. All natural processes, such as repetitive lumbar movement during cyclic lifting, show unique recurrent behaviour. Moreover, the recurrence of states, in the meaning that states are arbitrarily close after some time, is a fundamental property of deterministic dynamical systems and is typical for nonlinear or chaotic systems [239, 316]. The recurrence of natural phenomena was discovered many years ago; for example, the recurrence phenomena in cosmic-ray intensity was described in 1939 [317]. Eckmann and colleagues introduced recurrence plots as a tool for visualizing the recurrence of states x_i in a phase space [318]. Usually, a phase space does not have a dimension (two or three) that allows it to be pictured, and higher dimensional phase spaces can be visualized only by projection into the two- or three-dimensional subspace [239, 319]. *M*-dimensional phase space trajectories can be visualised and investigated through their two-dimensional representation in an $RP_{i,j}$. In the $RP_{i,j}$, the recurrence of a phase space state at time *i* at the subsequent time *j* is marked within a two-dimensional squared matrix with dots signifying ones and zeros (black-and-white dots in the plot), where both axes are time axes [239]. The fixed time period between *i* and *j* is called delay, and the number of *m* dimensions in the phase space is called embedding dimension [316]. Examples of the reconstructed phase space trajectory of a signal in a two-dimensional phase space and the corresponding RP_{ij} are presented in Figure 14**.** Mathematically, the $RP_{i,j}$ is expressed as

$$R_{i,j} = \Theta(\varepsilon_i - ||x_i - x_j||), \quad x_i \in \Re_m, \quad i,j = 1 \ldots N,$$

with N being the number of considered states x_i, ε_i the threshold distance (size of the neighbourhood or maximal distance at which points in the phase space are considered near or recurrent), $||\cdot||$ a norm, and $\Theta(\cdot)$ the Heaviside function [239, 315]. An RP_{ij} contains distinct structures. Closer inspection of the RPs reveals small-scale structures (the texture), which are single dots, and diagonal lines as well as vertical and horizontal lines (the combination of the latter two obviously forms rectangular clusters of

recurrence points) [239]. The presence of single, isolated recurrence points indicates rare states, states that do not persist, or states that fluctuate but do not indicate chance or noise [320]. Diagonal lines expressed as

$$R_{i+k,j+k} = 1, k = 1 \dots l$$

(where l is the length of the diagonal line) indicate segments running parallel to other segments (i.e., phase space trajectories that visit the same region of the phase space at different times) [239]. The diagonal line's length is determined by the duration of time that the phase space trajectories visit the same region of the phase space [243]. The predetermined threshold distance defines which distance around a phase space trajectory is considered the same region of the phase space. The direction of these diagonal structures can differ. Diagonal lines parallel to the line of identity (LOI) represent the parallel running of trajectories for the same time evolution. The diagonal structures perpendicular to the LOI represent the parallel running with contrary times [239]. Vertical and horizontal lines, expressed as

$$R_{i,j+k} = 1, k = 1 \dots v$$

where v is the length of the vertical line, indicate segments in which the phase space trajectory does not change or changes slowly [239, 243]. Figure 15 shows two sample RPs derived from a sinusoid signal and pink noise. The RP reconstructed from the sinusoid signals shows long diagonal lines indicating that the phase space trajectory of the underlying signal runs in parallel for prolonged times. The RP reconstructed from the pink noise shows single isolated points indicating an uncorrelated random process. The single points cluster around certain areas of the RP, indicating that the process is not entirely random but is influenced by a drift.

Sample entropy (SaEn) is a measure of complexity. It is used for assessing the complexity of a physiological time-series signal, thereby for diagnosing diseased states [244]. Entropy is the rate of information production and of chaos. Sample entropy calculations examine time series for similar epochs. An epoch is a phase in the development of a time series signal [315]. More frequent and similar epochs in a time series lead to lower values of SaEn and indicate that signals are less complex compared to signals containing higher values of SaEn [244]. A sinusoid signal contains a larger number of frequent and similar epochs compared to a signal consisting of pink noise; consequently, it has a lower value of entropy.

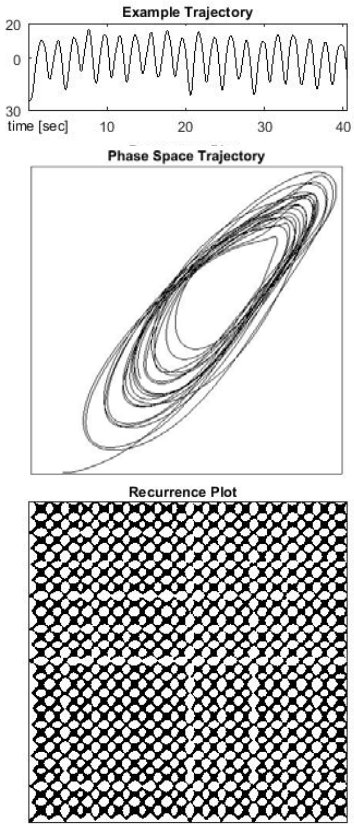


Figure 14. Reconstruction of movement data in a phase space and recurrence plot *Top:* Example trajectory (lumbar angle of one participant during a repetitive movement test, measured in Study III of this thesis). *Middle:* The reconstructed two-dimensional phase space trajectory. One point in the phase space is reconstructed from points from the movement data: the original point and a point at a defined distance (the delay) from the original point. The number of points used to reconstruct one point in the phase space is defined by its embedding dimension (2 in this example). Close points in the phase space form recurrent points in the recurrence plot. *Bottom:* The recurrence plot derived from the reconstructed phase space.

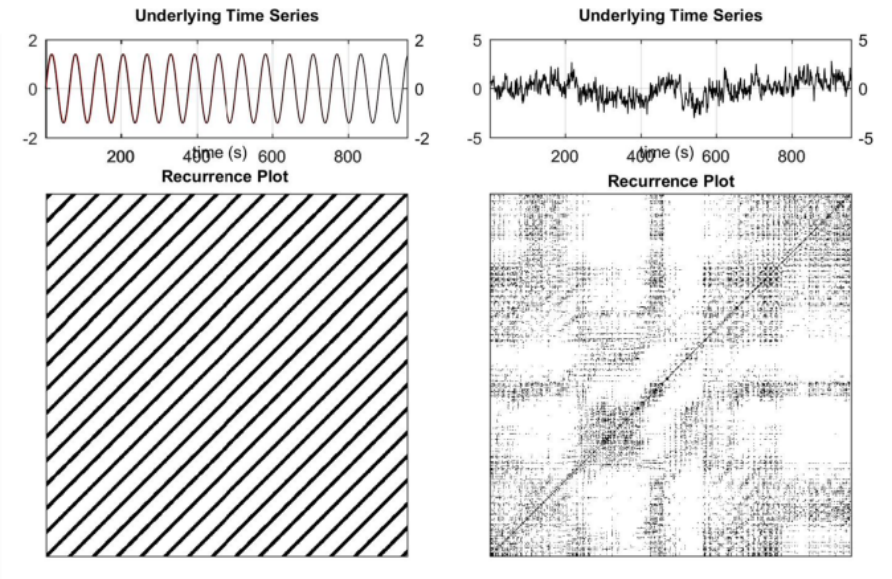


Figure 15. Example recurrence plots reconstructed from a sinosodial signal and pink noise. Left: Recurrence plot reconstructed from a sinusoid signal. Right: Recurrence plot reconstructed from pink noise.

Figure 16 shows sample angular displacement data from Study III: one painfree participant with high movement control and one participant with LBP and low movement control.

Table 11 describes the input parameters used to calculate the nonlinear indices of lumbopelvic kinematics in Studies I-V.

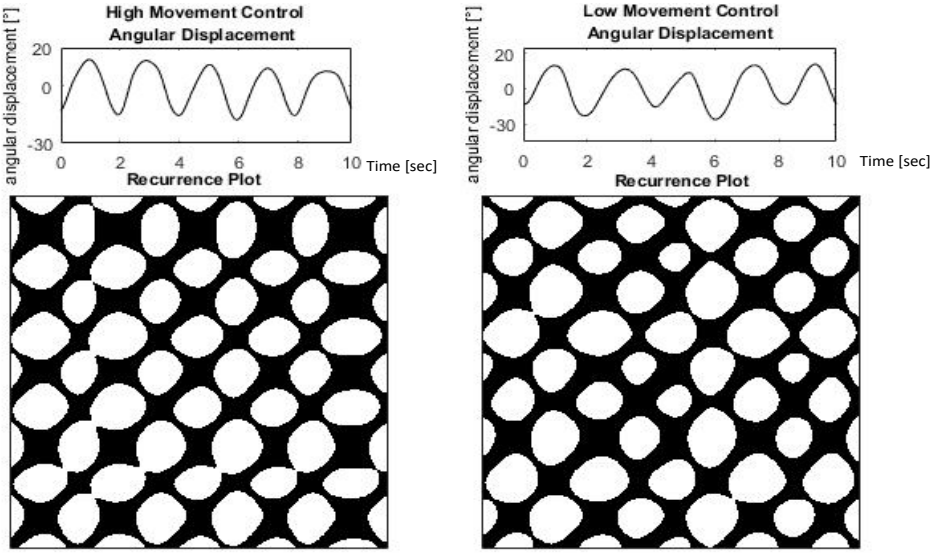


Figure 16. Study III: Angular displacement time series and recurrence plots. The left column shows a participant without LBP and high movement control, the right column a participant with high-intensity LBP and low movement control. Lumbar movement of the participant with high-intensity LBP and low movement control is less deterministic, indicated by the smaller amount of recurrent points that run in parallel to the line of identity.

Table 11. Input parameters for recurrence quantification analysis and sample entropy; σ = standard deviation of the underlying time series; l min = minimal length of diagonal line; PUB = Pick Up a Box; RFE = repeated flexion and extension. In all studies the Euclidian distance was used when determining neighbouring points and the tolerance was set at 1.3*σ for the RQA and 0.2*σ for the sample entropy calculation

				Recurrence Quantification Analysis					Sample Entropy	
				Embedding Dimension			l min	Tolerance	Embedding Dimension	
Study	**Test**	**Frequency**	**Filter**	**Delay**		**Distance**				**Tolerance**
Study II										
Angular Displacement	PUB	50	2nd, 6Hz Cutoff	15	4	Euclidian	20	1.3*σ	2	0.2*σ
Angular Velocity		50	—	14	4	Euclidian	20	1.3*σ	2	0.2*σ
Angular Acceleration		50	—	13	4	Euclidian	20	1.3*σ	2	0.2*σ
	RFE							1.3*σ		
Angular Displacement		50	2nd, 6Hz Cutoff	19	4	Euclidian	20	1.3*σ		
Angular Velocity		50	—	13	4	Euclidian	20	1.3*σ		
Angular Acceleration		50	—	14	4	Euclidian	20	1.3*σ		
Study III	PUB							1.3*σ		
Angular Displacement		200	2nd, 1Hz Cutoff	37	2	Euclidian	20	1.3*σ		
Angular Velocity		200	—	16	2	Euclidian	50	1.3*σ		
Angular Acceleration		200	—	9	2	Euclidian	20	1.3*σ		
Study IV	RFE							1.3*σ		
Angular Displacement		200	2nd, 6Hz Cutoff	35	2	Euclidian	150	1.3*σ	2	0.2*σ
Angular Velocity		200	—	21	2	Euclidian	150	1.3*σ	2	0.2*σ
Study V	PUB							1.3*σ		
Angular Displacement		50	2nd, 6Hz Cutoff	15	4	Euclidian	150	1.3*σ		
Angular Velocity		50	—	14	4	Euclidian	150	1.3*σ		

10.3 Study III Final model of each outcome

Table 12 Study III Final model of each outcome Abbreviations: 95% CI – 95 % confidence interval; AA – angular acceleration; AD – angular displacement; AV – angular velocity; BMI – body mass index; DET – determinism ; LB – lower bound; LBP – low back pain intensity; LS – lumbar spine; PS – physical stress at work; REC – recurrence rate; ROM – range of motion; UB – upper bound. The

Test & Variable	Covariate	Point Estimation	95% CI LB	95% CI UB	p-value
Sitting Knee Extension					
ROM LS (°)	LBP	0.3	-0.2	0.9	0.24
	Gender(Female)	0.8	-1.3	2.9	0.44
	BMI	-0.2	-0.4	0.0	0.05*
	LBP:Gender(Female)	-0.5	-1.3	0.2	0.14
Waiters Bow					
Log Ratio LS/Hip ROM	LBP	0.0	-0.1	0.2	0.84
	Gender(Female)	-0.1	-0.6	0.5	0.78
	BMI	0.1	0.0	0.1	0.03*
	PS	-0.4	-0.7	0.0	0.04*
	LBP:Gender(Female)	0.1	-0.1	-0.1	0.16
Pick Up the Box					
REC AD	LBP	0.11	-0.05	0.26	0.18
DET AD	LBP	-0.06	-0.11	-0.02	0.01*
	Age	-0.01	-0.01	0.00	0.11
REC AV	LBP	-0.25	-0.46	-0.03	0.03*
DET AV	LBP	-3.31	-6.21	-0.01	0.05*
	Gender(Female)	-2.22	-4.25	.-0.02	0.03*
	BMI	-0.66	-1.11	-0.25	0.45
	NRS:BMI	0.11	-0.02	0.24	0.11
REC AA	LBP	-2.03	-3.87	-0.20	0.03*
	Gender(Female)	-2.35	-5.10	0.47	0.09
	Age	-0.05	-0.16	0.06	0.40
	LBP:Gender(Female)	1.14	0.17	2.12	0.02*
	LBP:Age	0.03	-0.01	0.07	0.14
DET AA	LBP	-0.86	-1.73	0.00	0.05*
	BMI	-0.59	-1.06	-0.13	0.01*

reference level for gender was defined as female. The point estimation for each covariate is the effect of a one point increase of the respective covariate on the variable. For gender it represents the effect of being female.

*indicates $p \leq 0.05$

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Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement

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ABSTRACT

Introduction: Assessment of movement dysfunctions commonly comprises trunk range of motion (ROM), movement or control impairment (MCI), repetitive movements (RM), and reposition error (RE). Inertial measurement unit (IMU)-systems could be used to quantify these movement dysfunctions in clinical settings. The aim of this study was to evaluate a novel IMU-system when assessing movement dysfunctions in terms of concurrent validity and reliability. *Methods:* The concurrent validity of the IMU-system was tested against an optoelectronic system with 22 participants. The reliability of 14 movement dysfunction tests were analysed using generalizability theory and coefficient of variation, measuring 24 participants in seven trials on two days. *Results:* The IMU-system provided valid estimates of trunk movement in the primary movement direction when compared to the optoelectronic system. Reliability varied across tests and variables. On average, ROM and RM were more reliable, compared to MCI and RE tests. *Discussion:* When compared to the optoelectronic system, the IMU-system is valid for estimates of trunk movement in the primary movement direction. Four ROM, two MCI, one RM, and one RE test were identified as reliable and should be studied further for inter-subject comparisons and monitoring changes after an intervention.

1. Introduction

Movement dysfunctions in patients suffering from diseases such as low back pain (LBP), stroke and Parkinson's disease can be clinically assessed by measuring their trunk range of motion (ROM) and their reaction to specific movement control tasks (Laird et al., 2014; Verheyden et al., 2007; Cole et al., 2010). Specifically, these assessments are comprised of (1) ROM (Laird et al., 2014), (2) movement control impairment (MCI) (Sahrmann, 2002; Luomajoki et al., 2007), (3) repetitive movement (RM) tests (Dideriksen et al., 2014), and (4) tests for proprioception deficits such as reposition error tests (RE) (Rausch Osthoff et al., 2015).

Optoelectronic measurement systems are accepted as gold-standards for non-invasive analysis of trunk movement within research settings (Cuesta-Vargas et al., 2010, McGinley et al., 2009). However they are not applicable in daily clinical practice due to their high cost, required installation space, specific marker placement and subsequent data capture, analysis and processing. These factors limit the analysis to some standard procedures, which cannot be extended to clinics (Wong and Wong, 2009). Alternative objective, valid, and reliable measurement systems are needed to allow clinicians to assess and monitor individual patient changes and compare between different population groups.

To overcome these limitations, new wireless movement analysis systems using body-worn sensors have recently been developed (e.g. Valedo® from Hocoma AG, ViMove from dorsaVi, or Reablo® from Corehab). These clinical systems comprise of multiple small

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light weight inertial measurement units (IMU) which measure the angular tilt and velocity of body segments with respect to magnetic fields and gravity (Roetenberg et al., 2007). By combining the output of multiple IMU's and post processing algorithms into an IMU-system it is possible to estimate joint angles in a non-invasive way.

Using concurrent validation, the output of an IMU system can be correlated to the gold-standard, whilst simultaneously measuring with both systems (Streiner and Norman, 2008). Recent research examined concurrent validity of a wired IMU system and found a high correlation to the gold-standard (Wong and Wong, 2009; Wong et al., 2007). However correlation studies between two systems should provide both a measure of random error, or precision, as well as accuracy of the devices in their units of measurement (e.g. degrees). (de Vet et al., 2006). In a systematic review of the literature, Cuesta-Vargas and colleagues found that IMU systems can be concurrent to optoelectronic analysis of trunk measurements, but the degree of concurrent-validity is specific to the IMU system and anatomical site (Cuesta-Vargas et al., 2010).

Reliable measures of trunk movement and control are needed to monitor individual changes over time and to compare between different individuals. Reliability concerns the degree to which repeated measures provide similar results (de Vet et al., 2006). Reliability is affected by interrater, intrasession, and intersession variability (Corriveau et al., 2000). Interrater variability is unlikely to be a concern for measurements with an IMU system, except for sensor placement. Variability of sensor placement can be minimised by using a standardised protocol (Ernst et al., 2013). Intra- and intersession variability depend on biological variability, hence they are test specific. Reliable tests can be identified by estimating the magnitude of intra- and intersession variability. Furthermore, recommendations can be made for the number of trials needed to be averaged from one or more sessions in order to improve reliability (Santos et al., 2008).

This study assesses concurrent validity of a novel wireless IMU system, by using an optoelectronic system as a gold standard. Second, it investigates the reliability of commonly used trunk movement and control tests, when measured with a wireless IMU system.

2. Methods

This study was divided into two sub-studies: A concurrent validity study (study V) and a reliability study (study R).

2.1. Participants

Twenty-two and twenty-four asymptomatic participants volunteered for studies V and R respectively. The participant's characteristics are presented in Table 1. Detailed exclusion criteria for both studies are described elsewhere (Schelldorfer et al., 2015). For study R, the sample size was calculated according to Walter et al. (1998). Twenty participants and five trials allow reliability estimations of 0.95 with a type I error of 0.05 and a type II error of 0.20. The studies were approved by the local ethics commission and participants provided their informed consent.

2.2. Marker and sensor placement

Four IMUs were placed on the right thigh (THI), over the sacrum (S2), and at the level of L1 (L1), and T1 (T1), as described elsewhere (Ernst et al., 2013; Schelldorfer et al., 2015). The IMUs were mounted on a plastic frame and attached to the skin with hydrogel tape (KCI Medical GmbH 8153 Rümlang, CH). Reflective markers were placed above and below every IMU with a third marker attached to the stiletto on the plastic frame. Thus it was possible to build virtual segments corresponding to the IMU plane, and to compare the two systems (Fig. 1). The IMU and optoelectronic systems were synchronised using digital signals generated from a Labjack U3® data acquisition device (Labjack Corporation, USA).

Table 1
Participant's demographics (mean ± standard deviation).

Study V	All (n = 22)	Women (n = 11)	Men (n = 11)
Age (years)	41.18 ± 11.14	38.27 ± 10.44	44.09 ± 11.53
Body mass index	22.99 ± 2.89	22.67 ± 3.02	23.32 ± 2.85
Study R	All (n = 24)	Women (n = 13)	Men (n = 11)
Age (years)	38.04 ± 11.21	37.77 ± 10.12	38.44 ± 12.60
Body mass index	22.93 ± 2.69	22.58 ± 3.12	23.44 ± 1.85

n: Number of participants.

2.3. Measurement systems and data processing

Trunk movements were measured by the IMU system in both studies and additionally with an optoelectronic motion capture system (VICON, Oxford UK) in study V. In study V, a fourth-order zero-phase low-pass Butterworth filter (6 Hz cut-off frequency) was used to filter the raw data of both systems. In study R, an eighth-order zero-phase low-pass Butterworth filter (6 Hz cut-off frequency) was used since we analysed acceleration and jerk, which are noisy measures and require smoothing to obtain interpretable estimates.

2.3.1. Optoelectronic system

The optoelectronic system consisted of twelve infrared cameras. Data was sampled at 200 Hz and processed using VICON Nexus® software. The coordinate system of each segment, defined by three reflective markers, was aligned to the coordinate system of the IMU. The difference signal between two segments was calculated and transformed into tilt/twist angles according to Crawford and colleagues (Crawford et al., 1999). We adopted the following sign convention: flexion, lateral flexion towards the right, and axial rotation towards the left were assigned positive values; movements in the opposite directions were assigned negative values. We termed the angle between the L1 and T1 segment "Thoracic Spine", the angle between S2 and L1 "Lumbar Spine," and the angle between thigh and S2 "Hip angle".

2.3.2. Inertial measurement units

The Valedo® system (Hocoma AG) is a professional medical system used for low back pain therapy. The Valedo IMU's contain a tri-axillar gyroscope, magnetometer, and accelerometer, as well as wireless antenna and signal processing unit. The specifications of the IMU's indicate they are able to record ±0.1° over a range of 360° around all axes (Valedo® User Manual, Hocoma AG). IMU sensor data was transmitted to a recording computer with a 200 Hz sampling frequency. Custom data acquisition and synchronisation software (Valedo® Research) was provided by Hocoma AG. The raw IMU sensor data was transformed into quaternions according to Madgwick and colleagues (Madgwick et al., 2010). The angular difference between two IMU's placed above the body segments was calculated and transformed into tilt/twist angles. A complete description of the data processing from raw data to tilt/twist angles is documented in Supplementary File 1.

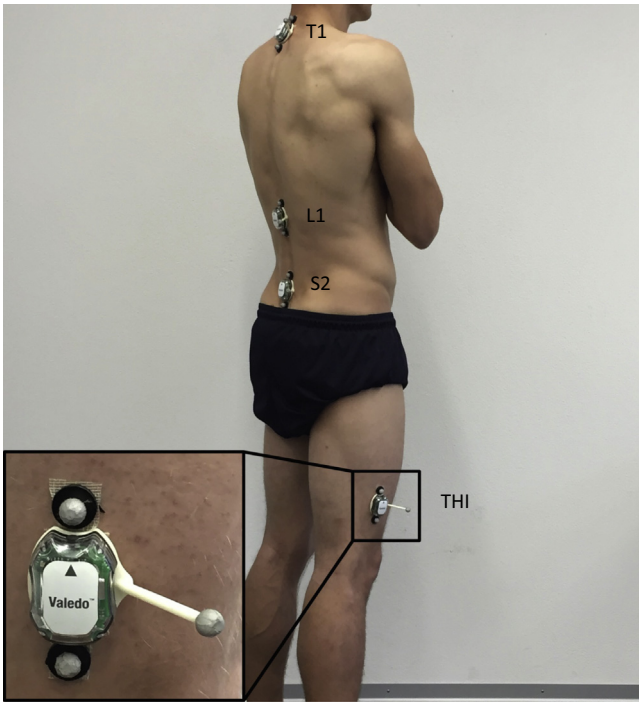

Fig. 1. Experimental setup: The THI, S2, L1, T1 IMU, and the reflective markers.

2.4. Procedures

2.4.1. Study V

Participants attended one measurement session and performed four ROM tests in randomized order, as described in Table 2. They were tutored by a video showing the correct movement. Additionally, they were instructed to move as far as possible at their preferred speed. Each test was performed three times.

2.4.2. Study R

Participants attended two identical measurement sessions, separated by a 1 week period. Both measurement sessions took place at the same time of day. All participants performed 14 tests, which were grouped into four categories according to their purposes: (1) ROM, (2) MCI, (3) RM and (4) RE. Test (1) measures the flexibility of the participant's spine within their comfort zone. Test (2) evaluates the participant's ability to control and differentiate movement between two body segments and to stabilize their spine. The former parameter was analysed by calculating the ratio of the ROM of the respective body segments, while the later was investigated using the ROM of the respective segment. Furthermore, the root mean squared jerk (RMSJ), as described by Slaboda et al. (2005), was calculated as indication of movement control. Test (3) measured the variability of angular displacement and acceleration during repeated movements. Variability was examined by calculating percentage of recurrence (%REC) and determinism (%DET) using recurrence quantification analysis (RQA) (Webber and Zbilut, 1994). Test (4) evaluates the participant's proprioceptive deficits within the spine, analysed using constant error (CE) (Rausch Osthoff et al., 2015).

Participants performed four ROM, six MCI, two RM, and two RE tests as described in Table 2. Each test was performed seven times, except for those in four point kneeling (4pk) which was reduced to 5 repetitions to minimise loading through their wrists. The order of the tests was randomized between participants but not between days.

2.5. Statistical analysis

2.5.1. Study V

The coefficient of determination (r^2), a measure of precision, and root mean squared error (RMSE), a measure of accuracy, were used to test the concurrent validity of the IMU system:

$$r^2 = 1 - \frac{\sum_i (y_i - \hat{y}_i)^2}{\sum_i (y_i - \bar{y})^2} \qquad \text{RMSE} = \sqrt{\frac{\sum_i (y_i - x_i)^2}{n}}$$

where x and y are the two time based movement signals, and $\hat{y}_i$ being the predicted value obtained by linear regression. The values of r^2 ranged from 0 to 1. A high value of r^2 implies that angles measured by IMUs and the optoelectronic system have the same characteristic. RMSE is the measure of the average difference between the two signals. Systematic differences between the systems were analysed using the Wilcoxon rank sum-test with p set at <0.05.

2.5.2. Study R

The generalizability theory (Brennan, 2001) with the design $p \times t \times d$ (*participants* × *trials* × *days*) was used as a framework to estimate reliability of trunk movement measures, based on the linear model.

$$X_{ptd} = \mu + v_p + v_t + v_d + v_{pt} + v_{pd} + v_{td} + v_{ptd}$$

with μ representing the global mean and v any one of the seven components.

The index of dependability Φ was calculated as:

$$\Phi = \frac{\sigma_p^2}{\sigma_p^2 + \frac{\sigma_t^2}{n_t} + \frac{\sigma_d^2}{n_d} + \frac{\sigma_{pt}^2}{n_t} + \frac{\sigma_{pd}^2}{n_d} + \frac{\sigma_{td}^2}{n_t n_d} + \frac{\sigma_{ptd}^2}{n_t n_d}}$$

with σ being the variance, and n the number of the corresponding component (with n_t, n_p, and n_d being the number of trials, participants, and days, respectively). Φ was interpreted as: <0.25 very low, 0.26–0.49 – low, 0.50–0.69 – moderate, 0.70–0.89 – high, and >0.90 – very high reliability (Carter et al., 2005). $\Phi \geqslant 0.70$ was interpreted as sufficient to compare between different individuals. Subsequently, Φ coefficients were calculated for alternative measurement strategies, where n_t was varied up to ten trials, and n_d varied across two days, which represent acceptable measurement strategies. Thereby, the number of required trials per day to achieve high reliability was evaluated.

The coefficient of variation (CV) (Hopkins, 2000) was calculated as

$$CV = \frac{\sigma_{diff}}{\sqrt{n_d} * \bar{x}} * 100$$

with $\bar{x}$ being the grand mean and σ_{diff} being the standard deviation of the differences between days and calculated from the mean of seven trials per day. The CV values were rated as follows: >10% not reliable, 6–10% adequately reliable and 5% highly reliable. CV's $\leqslant$ 10% were construed as sufficient to monitor changes over time (Suni et al., 2014).

The diagnostic value of a variable was assessed by Φ whereas the ability to detect changes over time was evaluated by the CV.

3. Results

3.1. Study V

In general, trunk movements in the sagittal plane were overestimated by the IMU system compared to the optoelectronic system (angular values between 1.3° and 6.5°). In contrast, frontal plane movements of the trunk were underestimated (angular values between 0.7° and 3.1°). Movements of the hip were measured

Table 2
Overview of the tests and variables for each test.

Test	Starting position	Movement	BS	Variable (unit)	Description of variable
ROM tests					
ROM flexion	Standing upright	Maximal flexion of the LS	LS	ROM_FLEX (°)	ROM LS
ROM extension	Standing upright	Maximal extension of the LS	LS	ROM_EXT (°)	ROM LS
ROM lateral flexion right	Standing upright	Maximal lateral flexion of the LS	LS	ROM_RIGHT (°)	ROM LS
ROM lateral flexion left	Standing upright	Maximal lateral flexion of the LS	LS	ROM_LEFT (°)	ROM LS
MCI tests (Luomajoki et al., 2007)					
pelvic tilt	Standing upright	Anterior pelvic tilt without moving the trunk or knees	LS	RATIO_PT	ROM LS/ROM TS
			TS	RMSJ_PT_LS (°/s^3) (Slaboda et al., 2005) RMSJ_PT_TS (°/s^3)	Smoothness of movement
Waiters bow	Standing upright	Hip flexion without moving the LS	LS	RATIO_WB	ROM LS/ ROM Hip
			Hip	RMSJ_WB_LS (°/s^3)	Smoothness of movement
				RMSJ_WB_Hip (°/s^3)	Smoothness of movement
Sitting knee extension	Sitting upright Hips at 90°	Knee extension without moving the LS	LS	ROM_SKE (°)	ROM LS
				RMSJ_SKE_LS (°/s^3)	Smoothness of movement
Rocking backwards	4pk	Hip flexion and shoulder extension without moving the LS	LS	RATIO_RB	ROM LS/ ROM Hip
			Hip	RMSJ_RB_LS (°/s^3)	Smoothness of movement
				RMSJ_RB_Hip (°/s^3)	Smoothness of movement
Rocking forwards	4pk	Hip extension and shoulder flexion without moving the LS	LS	RATIO_RF	ROM LS/ROM Hip
			Hip	RMSJ_RF_LS (°/s^3)	Smoothness of movement
				RMSJ_RF_Hip (°/s^3)	Smoothness of movement
Prone knee bend	Lying prone	Knee flexion without moving the LS	LS	ROM_PKB (°)	ROM LS
				RMSJ_PKB_LS (°/s^3)	Smoothness of movement
RM tests (Dideriksen et al., 2014)					
Picking up a box	Standing upright	Lifting a box (5% body weight) five times in a row at 60bpm	LS	%REC_PU_AD, %DET_ PU_AD (%) %REC_ PU_AA, %DET_ PU_AA (%)	Percentage of recurrence points within a recurrence plot (%REC)
Flexion and extension	Sitting upright Hips at 60°	Repeated flexion and extension of the trunk, five times in a row at 80bpm	LS	%REC_ FE_AD, %DET_ FE_AD (%) %REC_ FE_AA, %DET_FE_AA (%)	And percentage of recurrence points forming diagonal line structures in this plot (%DET) (Webber and Zbilut, 1994; Marwan et al., 2002; Rissanen et al., 2008)
RE Tests (Rausch Osthoff et al., 2015)					
Reposition Error Sitting	Sitting upright Hips at 60°	Flexion of the trunk and reproducing the starting position	LS	CE_SIT (°)	Angular difference between starting and final position
Reposition Error 4pk	4pk	Extension of the LS and reproducing the starting position	LS	CE_4PK (°)	Angular difference between starting and final position

4pk: four point kneeling; %DET: percentage of determinism; %REC: percentage of recurrence; AA = angular acceleration; AD: angular displacement; bpm: beats per minute; BS: Body segment; CE: constant error; EXT: Extension; FE: Flexion and Extension; FLEX: Flexion; LS: lumbar spine; MCI: Movement control impairment; PKB: prone knee bend; PT: Pelvic Tilt; PU: Picking Up a Box; RB: rocking backwards ;RE: Reposition Error; RF: rocking forwards; RM: repetitive movement; RMSJ: root mean squared jerk; ROM: range of motion; SKE sitting knee extension; SIT: sitting; TS: Thoracic Spine; WB: waiters bow.

almost equally with both systems. A summary of the results is presented in Table 3.

No significant systematic differences were found in the primary movement direction, except for sagittal and frontal plane movement of the thoracic spine (flexion and lateral flexion to the right).

The measurement systems showed acceptable agreement and small measurement errors in the primary movement direction. The r^2 coefficients ranged between 0.94 and 0.99, except for hip movement during the lateral flexion tests (0.85–0.87) and the RMSE ranged between 1.1° and 6.8°. Flexion of the lumbar spine and the hip, as well as lateral flexion of the thoracic and lumbar spine, revealed very high agreement with an r^2 coefficient of 0.99 and RMSE ranging between 1.8° and 6.1°. In the non-primary movement directions, r^2 coefficients were lower (0.36–0.87) while RMSE were similar (1.2°–6.8°) compared to the primary movement direction (Supplementary File 2).

3.2. Study R

Table 4 summarises the grand mean, Φ-coefficients, and the number of trials averaged from one or two measurement days which are needed to gain $\Phi \geqslant 0.70$, and the CV for each variable. On average, ROM and RM tests needed a smaller number of trials to reach high reliability and had smaller CVs compared to MCI and RE tests.

Measured values from single trial tests of trunk ROM revealed high to very high reliability except for extension of the lumbar spine. All CVs were smaller than 10%. The MCI tests differed in their reliability with Φ-coefficients of a single measurement ranging from low to high, and CVs from 8 to 22%. The RM tests showed CVs smaller than 10%, with the "Picking up a Box" test being more reliable than the "Flexion and Extension" test. The RE tests showed a respectively low reliability for a single measurement with CVs greater than 10%.

4. Discussion

The main findings of the present study were that the use of a wireless IMU system is a valid alternative to measure trunk movements in the primary movement direction when compared to the golden standard (i.e. an optoelectronic system). Secondly, on average, the ROM and RM tests needed a smaller number of repeated trials to reach high reliability and had smaller CVs when compared to the MCI and RE tests.

4.1. Study V

The measured ROM falls well within the range of previously published results, although comparability is hampered by a large variety of measurement approaches, including measurement systems and participants selection (Laird et al., 2014). Both our optoelectronic and IMU systems measured similar ROM, whilst sagittal plane movement was slightly overestimated, and frontal plane movement underestimated, by the IMU systems.

This study showed that trunk ROM in the primary movement direction can be accurately measured by using a wireless IMU system; however, the system appears less valid for assessing movements in non-primary directions. Although RMSE were similar in magnitude compared to the primary movement direction, they were higher relative to the total ROM. The agreement could be affected by the noise, and limited resolution of the IMU system, a nonlinear correlation between both systems, and constraints on mathematical calculations.

The present study improves upon previous work (Ha et al., 2013; Wong and Wong, 2009) with a more detailed analysis of ROM measures which includes thoracic spine and hip ROM. Furthermore, the concurrent validity of the novel wireless IMU system compares well to other studies validating different IMU systems against a gold-standard (Dunne et al., 2008; Ha et al., 2013; Wong and Wong, 2008, 2009).

4.2. Study R

The index of dependability Φ of a single trial varied across different tests and variables, ranging from 0.19 to 0.90. The CV varied considerably as well, ranging from <1% to 37%. Reliability can be improved by increasing the number of trials/days and using the mean value. While, for some variables, averaging over days affected reliability more than averaging over trials on one day, this is not necessarily a practical solution, especially in clinical settings. If one attempts to increase the number of trials, care should be taken that a learning-effect or fatigue does not influence the participants' performance (Santos et al., 2008).

4.2.1. Range of motion

Three out of the four lumbar ROM variables reached high reliability with a single trial on one day, whereas the extension ROM only had moderate reliability. Averaging two single trials over two days increased the reliability of ROM extension more than averaging several trials on one day, indicating that it is affected more by sources of variance between days rather than within one day. The decreased reliability of ROM extension could be explained by biological variability between days, the test-setup, or the slightly lower concurrent validity of the IMU system (Table 3).

The low CVs (3–9%) indicate high reliability for measuring changes in ROM over time. These results are in accordance with other studies reporting high reliability of ROM measures (Al Zoubi and Preuss, 2013). The measured ROM is almost identical to study V and within the range of previously published results (Laird et al., 2014).

4.2.2. Movement control impairment

The MCI tests differed in their reliability. "Waiters Bow" and "Sitting Knee Extension" reached high reliability when averaging a maximum of six trials on one day, or two trials on two days. The magnitude of the between-day variance is also shown by the CV, ranging between 8% and 22%. Nonetheless, the mean ROM in "Sitting Knee Extension" was approaching zero, with about 25% of participants moving into extension, hampering the interpretation of the CV (22%) for this variable. "Pelvic Tilt", "Rocking Forwards", "Rocking Backwards," and "Prone Knee Bend" showed little to moderate reliability. The reliability might be affected by the complexity or the standardisation of the MCI tests or because segment movement ranges, duration, and speed were not controlled. Standardising the MCI tests for one of these factors might decrease within-day and between-days variance.

Our results are somewhat contradictory in regard to previous research, where the reliability of MCI tests was reported as substantial based on a dichotomous variable (positive or negative indication) (Luomajoki et al., 2007). Although a growing body of research investigates MCI of the trunk and hip (Luomajoki et al., 2007; Saner et al., 2015), no normative values have been published aside from this study. Additionally, the different approaches to quantify MCI tests make it difficult to compare our results.

4.2.3. Repeated movement tests

The "Picking Up a Box" test had high reliability by averaging a maximum of four trials on one measurement day, with low CVs ($\leqslant$3%). Our descriptive results for %DET of angular displacement

Table 3
Study V results for trunk range of motion measures, primary movement direction.

	ROM thoracic spine (°)				ROM lumbar spine (°)				ROM Hip (°)			
	IMU system, ° Mean ± SD	Optoelectronic system, ° Mean ± SD	r^2 Mean ± SD	RMSE, ° Mean ± SD	IMU System, ° Mean ± SD	Optoelectronic System, ° Mean ± SD	r^2 Mean ± SD	RMSE, ° Mean ± SD	IMU System. ° Mean ± SD	Optoelectronic System, ° Mean ± SD	r^2 Mean ± SD	RMSE, ° Mean ± SD
ROM flexion	36.2 ± 11.9*	29.7 ± 10.9*	0.95 ± 0.04	5.8 ± 2.0	53.3 ± 10.9	50.71 ± 9.5	0.99 ± 0.01	4.1 ± 1.8	77.4 ± 15.3	77.1 ± 14.2	0.99 ± 0.01	6.1 ± 2.7
ROM extension	22.2 ± 9.9	18.9 ± 9.9	0.94 ± 0.09	5.9 ± 3.3	16.6 ± 10.5	15.3 ± 8.4	0.97 ± 0.05	4.4 ± 2.2	13.7 ± 5.8	14.8 ± 5.8	0.94 ± 0.09	5.6 ± 4.1
ROM lateral flexion right	31.9 ± 5.1*	35.0 ± 6.1*	0.99 ± 0.01	2.8 ± 1.4	22.8 ± 5.1	23.7 ± 5.1	0.99 ± 0.01	1.8 ± .0	7.3 ±4.3	7.3 ± 4.9	0.87 ± 0.21	1.1 ± .7
ROM lateral flexion left	32.6 ± 9.2	34.7 ± 10.3	0.99 ± 0.03	2.6 ± 2.0	22.2 ± 5.7	22.9 ± 6.5	0.99 ± 0.01	1.9 ± 1.3	6.8 ± 3.0	6.9 ± 3.4	0.85 ± 0.20	1.1 ± .7

IMU: inertial measurement units; r^2: *R*-squared; RMSE: root mean squared error; ROM: range of motion; SD: standard deviation.
* A significant systematic difference between the two systems.

are comparable with previous research (Dideriksen et al., 2014), which did not report reliability of their measures.

The "Flexion and Extension" test showed lower Φ-values, whilst the CVs were also small (⩽6%). "Picking Up a Box" is predominantly performed by flexing the spine and hips, while the second test is based on flexion and extension. In this study, measures of extension were less reliable and had lower concurrent validity, which might explain the lower Φ values. Both tests were highly standardised, possibly explaining the small standard deviations of these variables.

4.2.4. Reposition error

Reposition error, CE (Rausch Osthoff et al., 2015), reached high reliability after averaging six trials on one day (4pk) or eight trials across two days (sitting). The CE can have positive and negative values and a score of zero implies a good performance. These characteristics result in an expected grand mean around zero and, therefore, huge CVs. Consequently, the CV should not be interpreted for these two variables. In such situations Φ gives a better indication of reliability. The magnitude of the measured RE is well within the range of previously published data on pain-free participants (Rausch Osthoff et al., 2015). Data on reliability of RE measures is discouraging. Several studies report poor reliability of RE tests, use an inadequate numbers of trials, or do not report reliability of their measures (Rausch Osthoff et al., 2015).

4.3. Limitations of this study

The IMU system is a valid tool when measuring flexion of the lumbar spine and hip, as well as lateral flexion of the thoracic and lumbar spine. On the other hand, measurements of thoracic spine flexion and hip lateral flexion should be viewed with caution. Some of the differences between the two systems can be characterised as errors in the optoelectronic system. These errors could be triggered by camera noise, limited sight of markers, or vibrations of the marker frame (Ehara et al., 1997). Additionally both systems are affected by skin surface artefacts caused by contraction of the muscles or prominent spinal processes (Yang et al., 2008).

The sample size was calculated for an Intraclass-Correlation-Coefficient model (Walter et al., 1998). We assume this to be appropriate as both models share similarities while generalizability theory is regarded as an expansion of classical reliability theory (Brennan, 2001). RMSJ was calculated as a measure of movement control that has been shown to be reliable and discriminative between populations (Slaboda et al., 2005). However, RMSJ is sensitive to movement duration, amplitude, and arrest (Hogan and Sternad, 2009). Other indices of movement control could be investigated in future studies. This study has focused on pain-free participants. Although reliability is affected by the heterogeneity of study populations (Lariviere et al., 2013), the inclusion of pain-free participants was reasonable to evaluate the usability of an IMU system to measure trunk kinematics.

4.4. Suggestions for future research

The evaluated wireless IMU system is appropriate as a more affordable alternative to an optoelectronic system within the demonstrated boundaries regarding secondary movement directions. The IMU system's concurrent validity might be enhanced by investigating the technical validity of the IMU components and subsequently improving these components.

Future studies should address reliability on different populations and assess diagnostic value and the ability to detect changes of the presented measures over time in more detail.

Table 4
Study R Results of Trunk Movement Measures: Reliability of a single measure, number of trials averaged on one or two days needed to achieve high reliability and coefficient of variation.

Test	Variable; Unit	Mean ± SD	Φ one trial	Number trials $\Phi > 0.7$ one day	Number trials $\Phi > 0.7$ two days	CV (%)
ROM tests						
ROM flexion	ROM_FLEX (°)	53.6 ± 9.6	0.80	1	1	3
ROM extension	ROM_EXT (°)	−17.5 ± 7.9	0.63	>10	1	9
ROM lateral flexion right	ROM_RIGHT (°)	−20.7 ± 7.3	0.90	1	1	3
ROM lateral flexion left	ROM_LEFT (°)	21.2 ± 6.8	0.90	1	1	3
MCI tests						
Pelvic tilt	RATIO_PT	.16 ± 0.1	0.27	>10	7	16
	RMSJ_PT_TS (°/s^3)	4.2 ± 3.1	0.27	>10	>10	15
	RMSJ_PT_LS (°/s^3)	72.5 ± 49.1	0.35	>10	>10	20
Waiters bow	RATIO_WB	.54 ± .44	0.77	1	1	10
	RMSJ_WB_LS (°/s^3)	48.8 ± 31.7	0.68	2	1	8
	RMSJ_WB_Hip (°/s^3)	61.7 ± 35.5	0.51	6	2	11
Sitting knee extension	ROM_SKE (°)	1.9 ± 2.8	0.68	2	1	22
	RMSJ_SKE_LS (°/s^3)	17.5 ± 8.9	0.62	3	1	8
Rocking backwards	RATIO_RB	.71 ± .43	0.38	>10	>10	18
	RMSJ_RB_LS (°/s^3)	29.3 ± 12.9	0.44	8	2	10
	RMSJ_RB_Hip (°/s^3)	28.4 ± 9.9	0.39	8	3	8
Rocking forward	RATIO_RF	1.52 ± 1.16	0.19	>10	>10	11
	RMSJ_RF_LS (°/s^3)	35.2 ± 20.9	0.73	1	1	9
	RMSJ_RF_Hip (°/s^3)	31.1 ± 12.4	0.31	>10	9	12
Prone knee bend	ROM_PKB (°)	−4.0 ± 2.7	0.44	>10	3	14
	RMSJ_PKB_LS (°/s^3)	24.9 ± 13.5	0.45	>10	8	14
RM tests						
Picking up a box	%REC_PU_AD (°)	0.15 ± 0.01	0.68	3	2	2
	%DET_PU_AD (°)	0.97 ± 0.01	0.51	3	2	<1
	%REC_PU_AA (°/s^2)	0.13 ± 0.01	0.63	4	2	3
	%DET_PU_AA (°/s^2)	0.74 ± 0.04	0.65	3	2	2
Flexion and extension	%REC_FE_AD (°)	0.13 ± 0.01	0.29	>10	>10	4
	%DET_FE_AD (°)	0.97 ± 0.01	0.64	5	3	<1
	%REC_FE_AA (°/s^2)	0.08 ± 0.01	0.24	>10	>10	4
	%DET_FE_AA (°/s^2)	0.66 ± 0.07	0.60	6	3	6
RE tests						
Reposition error sitting	CE_SIT (°)	−.94 ± 1.4	0.19	>10	8	37
Reposition error 4pk	CE_4PK (°)	1.6 ± 1.8	0.30	6	4	22

4pk: four point kneeling; Φ: index of dependability; %DET: percentage of determinism; %REC: percentage of recurrence; AA = angular acceleration; AD: angular displacement; CE: constant error; CV: Coefficient of variation; EXT: Extension; FE: Flexion and Extension; FLEX: Flexion; LS: lumbar spine; MCI: Movement control impairment; PKB: prone knee bend; PT: Pelvic Tilt; PU: Picking Up a Box; RB: rocking backwards ;RE: Reposition Error; RF: rocking forwards; RM: repetitive movement; RMSJ: root mean squared jerk; ROM: range of motion; SKE sitting knee extension; SIT: sitting; SD: Standard deviation; TS: Thoracic Spine; WB: waiters bow.

Differences between populations and treatment effects of interventions aiming at improving movement control have to be investigated. Measures of RQA in repeated movement tests are highly dependent on the input parameters (Rissanen et al., 2008; Webber and Zbilut, 1994). Other choices for input parameters, apart from the ones used in our study (Table 5), are possible, and optimal input parameters have to be investigated in future studies.

4.5. Clinical implications and recommendations

Clinicians commonly use range of motion and movement control tests of the trunk and hip to assist in identifying patterns of dysfunction and to monitor change (Laird et al., 2014). This paper presents a measurement tool which enables the clinicians to do this objectively. To identify dysfunctions and changes in performance, high reliability is important. Based on our results, we recommend the use of four ROM tests, selected MCI tests ("Waiters Bow" and "Sitting Knee Extension"), RE in 4pk, and "Picking up a Box" for RM, using an adequate number of trials for each test (Table 4).

5. Conclusion

The usage of a wireless IMU system led to valid estimates of trunk movement in the primary movement directions. A number of tests to assess movement dysfunctions and their corresponding variables were identified as reliable and should be studied further for intersubject comparisons and monitoring changes after an intervention.

Table 5
Input parameters used in recurrence quantification analysis.

Test	Delay	Embedding dimension
Picking up a box		
Angular displacement	15	4
Angular acceleration	13	4
Flexion and extension		
Angular displacement	19	4
Angular acceleration	14	4

Conflict of interest

The authors declare that they have no conflict of interest.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jelekin.2015.06.001.

References

Al Zoubi FM, Preuss RA. Reliability of a measure of total lumbar spine range of motion in individuals with low back pain. J Appl Biomech 2013;29(6):670–7.

Brennan RL. Generalizability theory. 2nd ed. New York: Springer; 2001.

Carter RE, Lubinsky J, Domholdt E. Rehabilitation research: principles and applications. 4th ed. St. Louis: Elsevier Saunders; 2005.

Cole MH, Silburn PA, Wood JM, Worringham CJ, Kerr GK. Falls in Parkinson's disease: kinematic evidence for impaired head and trunk control. Mov Disord 2010;35(14):2369–78.

Corriveau H, Hébert R, Prince F, Raiche M. Intrasession reliability of the «center of pressure minus center of mass» variable for postural control in the healthy elderly. Arch Phys Med Rehabil 2000;81(1):45–8.

Crawford NR, Yamaguchi GT, Dickman CA. A new technique for determining 3-D joint angles: the tilt/twist method. Clin Biomech 1999;14(3):153–65.

Cuesta-Vargas AI, Galàn-Mercant A, Williams JM. The use inertial sensors system for human motion analysis. Phys Ther Rev 2010;15(6):105–10.

de Vet HC, Terwee CB, Knol DL, Bouter LM. When to use agreement versus reliability measures. J Clin Epidemiol 2006;59(10):1033–9.

Dideriksen JL, Gizzi L, Petzke F, Falla D. Deterministic accessory spinal movement in functional tasks characterizes individuals with low back pain. Clin Neurophysiol 2014;125(8):1663–8.

Dunne LE, Walsh P, Herman S, Smyth B, Caulfield B. Wearable monitoring of seated spinal posture. IEEE Tran Biomed Circuits Syst 2008;2(2):97–105.

Ehara Y, Fujimoto H, Miyazaki S, Mochimaru M, Tanaka S, Yamamoto S. Comparison of the performance of 3D camera systems II. Gait Posture 1997;5(3):251–5.

Ernst M, Rast F, Bauer C, Marcar V, Kool J. Determination of thoracic and lumbar spinal processes by their percentage position between C7 and the PSIS level. BMC Res Notes 2013;6:58.

Ha TH, Saber-Sheikh K, Moore AP, Jones MP. Measurement of lumbar spine range of movement and coupled motion using inertial sensors - a protocol validity study. Man Ther 2013;18(1):87–91.

Hocoma AG. Valedo®Motion user manual. 2nd ed. Volketswil: Hocoma AG; 2011.

Hogan N, Sternad D. Sensitivity of smoothness measures to movement duration, amplitude, and arrests. J Mot Behav 2009;41(6):529–34.

Hopkins WG. Measures of reliability in sports medicine and science. Sports Med 2000;30(1):1–15.

Laird RA, Gilbert J, Kent P, Keating JL. Comparing lumbo-pelvic kinematics in people with and without back pain: a systematic review and meta-analysis. BMC Musculoskelet Disord 2014;15:229.

Lariviere C, Mecheri H, Shahvarpour A, Gagnon D, Shirazi-Adl A. Criterion validity and between-day reliability of an inertial-sensor-based trunk postural stability test during unstable sitting. J Electromyogr Kinesiol 2013;23(4):899–907.

Luomajoki H, Kool J, de Bruin ED, Airaksinen O. Reliability of movement control tests in the lumbar spine. BMC Musculoskelet Disord 2007;8:90.

Madgwick S, Vaidyanathan R, Harrison A. An Efficient Orientation Filter for IMU and MARG Sensor Arrays. University of Bristol: Department of Mechanical Engineering; 2010.

Marwan N, Wessel U, Meyerfeldt A, Schirdewan A, Kurths J. Recurrence plot based measures of complexity and its application to heart rate variability data. Phys Rev E 2002;66:026702.

McGinley JL, Baker R, Wolfe R, Morris ME. The reliability of three-dimensional kinematic gait measurements: a systematic review. Gait Posture 2009;29(3):360–9.

Rausch Osthoff AK, Ernst MJ, Rast FM, Mauz D, Graf ES, Kool J, et al. Measuring lumbar reposition accuracy in patients with unspecific low back pain: systematic review and meta-analysis. Spine 2015;40(2):E97–E111.

Rissanen SM, Kankaanpaa M, Meigal A, Tarvainen MP, Nuutinen J, Tarkka IM, et al. Surface EMG and acceleration signals in Parkinson's disease: feature extraction and cluster analysis. Med Biol Eng Comput 2008;46(9):849–58.

Roetenberg D, Baten CTM, Veltink PH. Estimating body segment orientation by applying inertial and magnetic sensing near ferromagnetic materials. IEEE Trans Neural Syst Rehabil Eng 2007;15(3):469–71.

Sahrmann S. Diagnosis and treatment of movement impairment syndromes. St. Louis, Missouri 63146 USA: Mosby; 2002.

Saner J, Kool J, Sieben JM, Luomajoki H, Bastianen CHG, De Bie RA. A tailored exercise program versus general exercise for a subgroup of patients with low back pain and movement control impairment: a randomised controlled trial with one-year follow-up. Man Ther 2015.

Santos BR, Delisle A, Lariviere C, Plamondon A, Imbeau D. Reliability of centre of pressure summary measures of postural steadiness in healthy young adults. Gait Posture 2008;27(3):408–15.

Schelldorfer S, Ernst MJ, Rast FM, Bauer CM, Meichtry A, Kool J. Low back pain and postural control, effects of task difficulty on centre of pressure and spinal kinematics. Gait Posture 2015;41(1):112–8.

Slaboda JC, Boston JR, Rudy TE, Lieber SJ, Rasetshwane DM. The use of splines to calculate jerk for a lifting task involving chronic lower back pain patients. IEEE Trans Neureal Syst Rehabil Eng 2005;13(3):406–14.

Streiner DL, Norman GR. Health measurement scales: a practical guide to their development and use. 4th ed. Oxford: OUP; 2008.

Suni J, Rinne M, Ruiz J. Retest repeatability of motor and musculoskeletal fitness tests for public health monitoring of adult populations. J Nov Physiother 2014;4:1.

Verheyden G, Nieuwboer A, Van de Winckel A, De Weerdt W. Clinical tools to measure trunk performance after stroke: a systematic review of the literature. Clin Rehabil 2007;21(5):387–94.

Walter SD, Eliasziw M, Donner A. Sample size and optimal designs for reliability studies. Stat Med 1998;17(1):101–10.

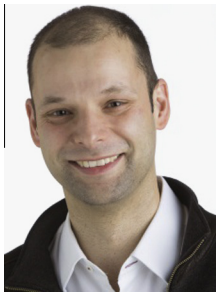
Webber Jr CL, Zbilut JP. Dynamical assessment of physiological systems and states using recurrence plot strategies. J Appl Physiol 1994;76(2):965–73.

Wong WY, Wong MS. Trunk posture monitoring with inertial sensors. Eur Spine J 2008;17(5):743–53.

Wong WY, Wong MS. Measurement of postural change in trunk movements using three sensor modules. IEEE Trans Instru Measure 2009;58(8):2737–42.

Wong WY, Wong MS, Lo KH. Clinical applications of sensors for human posture and movement analysis: a review. Prosthet Orthot Int 2007;31(1):62–75.

Yang Z, Ma HT, Wang D, Lee R. Error analysis on spinal motion measurement using skin mounted sensors. Conf Proc IEEE Eng Med Biol Soc 2008:4740–3.



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Technical and measurement report

Reliability of lumbar movement dysfunction tests for chronic low back pain patients

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ABSTRACT

Assessment of lumbar movement dysfunction commonly comprises trunk range of motion (ROM), movement or control impairment (MCI), and reposition error (RE). Those assessments are typically based on visual observation. Consequently it is not possible to reliably quantify back movements for inter-subject comparisons, or for monitoring changes before and after an intervention. Inertial measurement unit (IMU)-systems could be used to quantify these movement dysfunctions in clinical settings. The aim of this study was to evaluate the reliability of movement dysfunction tests when measured with a novel IMU-system. The reliability of eleven movement dysfunction tests (four ROM, six MCI and one RE tests) were analysed using generalizability-theory and minimal detectable change, measuring 21 chronic low back pain patients in seven trials on two days. Reliability varied across tests and variables. Four ROM and selected MCI tests and variables were identified as reliable. On average, ROM test were more reliable, compared to MCI and RE tests. An attempt should be made to improve the reliability of MCI and RE measures, for example through better standardizations. Subsequently these measures should be studied further for intersubject comparisons and monitoring changes after an intervention.

1. Background and purpose

Low back pain (LBP) is a common disorder with a lifetime prevalence as high as 84% and a high probability of recurrence (Airaksinen et al., 2006; Costa Lda et al., 2009). Contemporary LBP triage systems propose that there is a large group of patients who present with movement dysfunctions (MD), which are a relevant and a provocative factor for ongoing pain (O'Sullivan, 2005; Vibe Fersum et al., 2009). Tests for MD are specifically comprised of 1) range of motion (ROM) (Laird et al., 2014), 2) movement control impairments (MCI) (Sahrmann, 2002; Luomajoki et al., 2007) and 3) tests for proprioception deficits such as reposition error tests (RE) (Rausch Osthoff et al., 2015). These tests typically consist of visual observation (Oesch et al., 2007) and do not quantify MD for diagnostic and outcome evaluation purposes (Seffinger et al., 2004; van Trijffel et al., 2005; May et al., 2006; Stochkendahl et al., 2006; Littlewood and May, 2007).

To overcome these limitations, wireless movement analysis systems using body-worn sensors have recently been developed (e.g. Valedo® from Hocoma AG, ViMove from dorsaVi, or Reablo® from Corehab). These clinical systems comprise of multiple small light weight inertial measurement units (IMU) (Roetenberg et al., 2007). By combining the output of multiple IMU's and post processing algorithms into an IMU-system it is possible to estimate joint angles in a non-invasive way. In a previous study one IMU-system, consisting of four IMUs, was found to be concurrently valid for measures of trunk and hip movement (Bauer et al., 2015).

One prerequisite for tests on MD is high reliability. Four ROM tests and two MCI tests were found to have high reliability, in an asymptomatic population, when measured with an IMU-system (Bauer et al., 2015). However reliability is dependent on the

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heterogeneity of the sample and is therefore only applicable for a population with a similar heterogeneity (de Vet et al., 2006). Data on reliability of MD tests, when measured with the IMU-system, in patient populations such as CLBP patients, is currently lacking. This study assesses the reliability of ROM, MCI, and RE tests, in a population with CLBP and gives recommendations for reliable measurement protocols.

2. Methods

2.1. Participants

Twenty-three CLBP patients were recruited from a rehabilitation centre. Participants were between 18 and 65 years old and were suffering from CLBP for more than twelve weeks. Exclusion criteria were serious pathologies such as non-healed fractures, anomalies, tumours, specific LBP with neurological signs (muscle weakness, sensation or reflex loss) and acute trauma. Participants had to be able to understand German. The regional ethics committee granted approval. All participants gave their written informed consent.

2.2. Measurement system

Four IMUs (Valedo®) were placed on the right thigh (THI), over the sacrum (S2), and at the level of L1 (L1), and T1 (T1), as described elsewhere (Ernst et al., 2013; Schelldorfer et al., 2015) (Fig. 1). The IMUs were mounted on a plastic frame and attached to the skin with hydrogel tape (KCI Medical GmbH 8153 Rümlang, CH). The IMU's contain a tri-axillar gyroscope, magnetometer, and accelerometer, as well as wireless antenna and signal processing unit. IMU sensor data were transmitted to a recording computer with Valedo® Research software, with a 50 Hz sampling frequency. The raw IMU sensor data was transformed into quaternions according to Madgwick et al. (2010). The angular difference between two IMU's placed above the body segments was calculated and transformed into tilt/twist angles (Crawford et al., 1999). A complete description of the data processing from raw data to tilt/twist angles is documented elsewhere (Bauer et al., 2015). Data processing and statistics were performed using Matlab™.

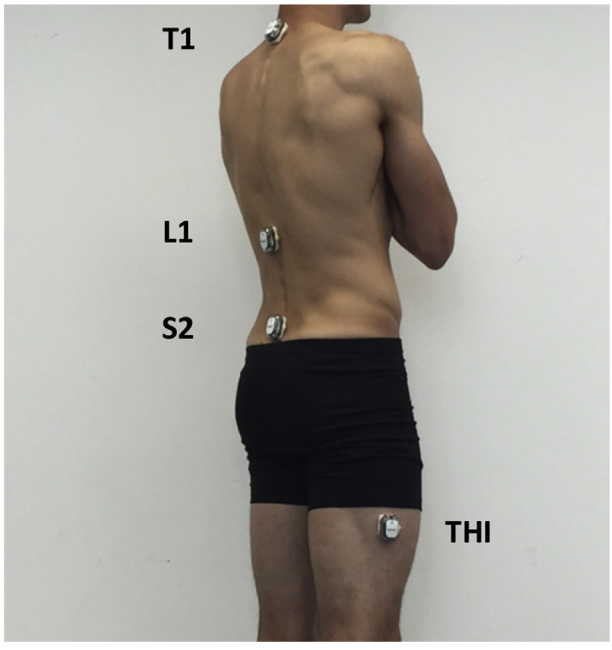

Fig. 1. Experimental setup: IMUs were placed on the right thigh (THI), and level of sacrum (S2), L1 (L1) and T1 (T1).

2.3. Test protocol

Participants attended two identical measurement sessions and performed eleven active movement tests twice within eight days. For retest, the IMU position was marked during the first measurement using a waterproof pen. Test and retest of a participant were conducted by the same examiner. Each session consisted of four ROM, six MCI and one RE tests (Table 2). The order of the tests was randomized between participants but not between sessions. Each test was repeated five times. It took a participant approximately 30 min to perform all tests with five repetitions. The participants did not, to the investigators knowledge, practice the tests between the IMU testing. They were instructed to not alter their routines while they participated in this study. Tests of ROM measure the flexibility of the participant's spine to the end of active range. Tests of MCI evaluate the participant's ability to differentiate movement between two body segments, to stabilize their spine and to move smoothly. These features were analysed by calculating the ratio of the ROM of the respective body segments, by measuring the ROM of the lumbar spine and by the root mean squared jerk (RMSJ). Jerk is defined as the rate of change of angular acceleration and quantifies smoothness of movement (Slaboda et al., 2005). Tests for RE evaluate the participant's proprioceptive deficits within the spine, analysed using absolute error (AE) and constant error (CE) (Rausch Osthoff et al., 2015). Prior to each test the participants received standardized oral instructions by one of the examiners and visual instructions in a video. In case of poor initial performance these instructions were repeated up to three times and the test was demonstrated by one examiner. If the participant was still performing the test incorrectly it was permitted. The participants were instructed to perform the tests at their own preferred speed. Detailed test descriptions and illustrations are provided in supplementary file.

2.4. Statistics

Generalizability theory was used to estimate reliability (Brennan, 2001), with the design $p \times t \times d$ (*participants* × *trials* × *days*) based on the linear model

$$X_{ptd} = \mu + \nu_p + \nu_t + \nu_d + \nu_{pt} + \nu_{pd} + \nu_{td} + \nu_{ptd}$$

with μ representing the global mean and ν any one of the seven components.

The index of dependability Φ was calculated as:

$$\Phi = \frac{\sigma_p^2}{\sigma_p^2 + \frac{\sigma_t^2}{n_t} + \frac{\sigma_d^2}{n_d} + \frac{\sigma_{pt}^2}{n_t} + \frac{\sigma_{pd}^2}{n_d} + \frac{\sigma_{td}^2}{n_t n_d} + \frac{\sigma_{ptd}^2}{n_t n_d}}$$

with σ being the variance, and n the number of the corresponding component (with n_t and n_d being the number of trials and days). N_t and n_d were equal to one to establish the reliability of a single trial. Φ was interpreted as: <0.25 very low, 0.26–0.49 – low, 0.50–0.69 – moderate, 0.70–0.89 – high, and >0.90 – very high reliability (Carter et al., 2005). Subsequently, Φ coefficients were calculated for alternative measurement strategies, where n_t was varied up to ten trials, and n_d varied across two days, which represent acceptable measurement strategies. Thereby, the number of required trials per day to achieve high reliability was evaluated. High reliability was interpreted as sufficient to compare between different individuals.

2.5. Minimal detectable change (MDC)

$$MDC = 1.96 \times \frac{\sigma_{diff}}{\sqrt{n_d}} \times \sqrt[2]{2}$$

with σ_{diff} being the standard deviation of the differences between days and was calculated from the means of five trials per day. The magnitude of MDC has to be interpreted individually for each variable and research question.

The diagnostic value of a variable was assessed by Φ whereas the ability to detect changes over time was evaluated by the MDC.

3. Results

In this study, 23 participants were recruited. Two participants did not participate on the second measurement day because their level of pain changed more than 2 points on the NRS between the two measurement days. Table 1 shows the descriptive data of the remaining 21 participants.

Table 2 summarizes the grand mean, Φ-coefficients, and the number of trials averaged from one or two measurement days which are needed to gain $\Phi \geq 0.70$and MDC for each variable. On average, ROM tests needed a smaller number of trials to reach high reliability, compared to MCI and RE tests.

Measured values from single trials of trunk ROM tests revealed high reliability. The minimal detectable change ranged between 3.7° and 6.5°.

Measures of MCI differed in their reliability with Φ-coefficients for single measures ranging between very low and very high (0.07–0.90). On average, measures of Pelvic Tilt, Waiters Bow and Prone Knee Bend needed fewer trials to reach high reliability than Sitting Knee Extension, Rocking Forward and Rocking Backwards.

Measures of RE had very low reliability, with Φ-coefficients for single measures ranging between 0.02 and 0.08. The MDC ranged between 2.4° and 3.3°.

4. Discussion

In this study we investigated the reliability of four ROM, six MCI and one RE test for the lumbar spine, in CLBP patients. On average, the ROM showed greater Φ compared to MCI and RE tests. This indicates that measures of ROM may be less affected by within-day and between-day variations, in a CLBP population.

The ROM tests reached high reliability with either a single or two trials measured on one day. The MDC ranged between 3.7° and 6.5°, indicating that any measured changes over time exceeding such values can be attributed to true change, and not measurement error. The increased MDC of ROM extension could be explained by biological variability between days, or the lower concurrent validity of the IMU-system, when measuring ROM extension (Bauer et al., 2015). A previous studies found high reliability for ROM tests in pain free participants (Bauer et al., 2015), indicating that participants from both populations can be ranked and assessed for changes over time. These results are in accordance with other studies reporting high reliability of ROM measures (Al Zoubi and Preuss, 2013).

The MCI tests differed in their reliability. The ratio of lumbar spine and hip and ROM measured in “Waiters Bow” and the RMSJ measured in “Pelvic Tilt”, “Waiters Bow” and “Prone Knee Bend” showed high reliability when averaging a maximum of three trials on one day, or two trials on two days. The MDC of those variables indicate that, compared to the grand mean, rather small changes over time could be attributed to real changes, not measurement error. “Sitting Knee Extension”, “Rocking Forwards”, and “Rocking Backwards” showed little to moderate reliability. These results are in line with previous research on pain-free participants (Bauer et al., 2015), except for “Sitting Knee Extension”, where the magnitude and variability of lumbar spine movement in the CLBP population were smaller compared to pain-free participants, possibly explaining the very low Φ. It was assumed that between-subject variance would generally be greater in CLBP patients compared to asymptomatic participants resulting in greater Φ, thus previously unreliable tests were included. The reliability may be affected by the complexity or the standardisation of the MCI tests, or because segment movement ranges, duration, timing, and speed were not controlled. Standardizing the MCI tests for one of these factors might decrease within-day and between-days variance. Tests with greater ROM may be less affected by noise, limited resolution of the IMU system, and constraints on mathematical calculations, possibly explaining why those tests are more reliable. Thus, it might be necessary to apply more accurate measurement tools to investigate tests with small ROM. However, the current study design is not able to distinguish between variability of test performance and measurement errors. Furthermore, between-day effects, such as learning or treatment effects and changes in pain might have affected test performance and thus reliability. A shorter period between the two measurement sessions might decrease between-days variance. Possibly MCI is an unstable phenomenon. Longitudinal studies investigating MCI might give further insight into whether it is a stable and relevant feature of CLBP.

These results are in line with previous research on pain-free participants (Bauer et al., 2015) but somewhat contradictory in regard to other previous research, where the reliability of MCI tests was reported as substantial based on a dichotomous variable (positive or negative indication) (Luomajoki et al., 2007). The methodological differences regarding the measurement systems might explain this discrepancy.

Reposition error was expressed as AE and CE (Rausch Osthoff et al., 2015), and showed low reliability even after averaging ten trials over two days. Data on reliability of RE measures is generally discouraging, with several studies reporting poor reliability of RE tests (Koumantakis et al., 2002; Rausch Osthoff et al., 2015).

4.1. Clinical implication

The application of the IMU system enables clinicians and researchers to objectively quantify movement dysfunctions, and to monitor longitudinal changes of movement dysfunctions associated with LBP. This may improve insight into the role of movement dysfunctions in LBP as well as LBP treatment efficacy.

5. Conclusion

Four tests of ROM and selected measures of MCI were identified as reliable, in a CLBP population. One RE test showed very low reliability. An attempt should be made to improve the reliability of MCI and RE measures, and then these measures should be studied

Table 1
Descriptive statistics of participants (mean ± SD).

Gender	n	Age	BMI	NRS	Oswestry
M	17	32.5 ± 10.5	25.4 ± 7.5	2.6 ± 1.1	7.8 ± 2.4
F	4	40.3 ± 18.8	23.2 ± 1.7	3.5 ± 2.1	8.5 ± 0.7

M – male, f – female, BMI – body mass index, NRS – average pain in the past two weeks, measured with a numeric rating scale (0 no pain–10 worst pain imaginable), Oswestry – Oswestry Disability Index Score (0–50 points).

Table 2
Trunk Movement Measures: Reliability of a single measure, number of trials averaged on one or two days needed to achieve high reliability and minimal detectable change.

Test	Variable; unit	Mean ± SD	Φ one trial	Number trials Φ >0.7 one day	Number trials Φ >0.7 two days	MDC
ROM tests						
ROM Flexion	ROM (°)	51.7 ± 9.2°	0.85	1	1	3.7°
ROM Extension	ROM (°)	−20.1 ± 15.4°	0.87	1	1	6.5°
ROM Lateral Flexion Right	ROM (°)	−18.3 ± 6.9°	0.69	2	1	4.6°
ROM Lateral Flexion Left	ROM (°)	18.3 ± 6.3°	0.68	2	1	3.9°
MCI tests						
Pelvic Tilt	Ratio TS/LS	0.8 ± 0.4	0.22	>10	>10	0.7
	RMSJ TS (°/s^3)	19.8 ± 12.6°/s^3	0.62	3	1	12.6°/s^3
	RMSJ LS (°/s^3)	98.7 ± 56.7°/s^3	0.60	3	1	42.4°/s^3
Waiters Bow	Ratio LS/Hip	0.7 ± 0.5	0.91	1	1	0.3
	RMSJ LS (°/s^3)	49.0 ± 30.2°/s^3	0.45	>10	2	17.8°/s^3
	RMSJ Hip (°/s^3)	61.4 ± 55.5°/s^3	0.60	8	1	20.2°/s^3
Sitting Knee Extension	ROM LS (°)	0.9 ± 1.3°	0.13	>10	>10	1.8°
	RMSJ LS (°/s^3)	18.0 ± 7.1°/s^3	0.42	8	3	7.3
Rocking Backwards	Ratio LS/Hip	1.2 ± 1.0	0.30	9	6	1.3
	RMSJ LS (°/s^3)	28.3 ± 13.6°/s^3	0.28	>10	>10	13.6°/s^3
	RMSJ Hip (°/s^3)	29.4 ± 13.7°/s^3	0.29	>10	>10	13.1°/s^3
Rocking Forward	Ratio LS/Hip	2.0 ± 1.4	0.36	8	4	0.9
	RMSJ LS (°/s^3)	31.0 ± 12.1°/s^3	0.15	>10	>10	14.7°/s^3
	RMSJ Hip (°/s^3)	26.7 ± 12.9°/s^3	0.07	>10	>10	9.7
Prone Knee Bend	ROM LS (°)	−1.2 ± 2.3°	0.11	>10	>10	2.8°
	RMSJ LS (°/s^3)	28.0 ± 12.7°/s^3	0.66	2	1	8.8°/s^3
RE tests						
Reposition Error sitting	CE (°)	−1.1 ± 1.5°	0.02	>10	>10	3.3°
	AE (°)	2.1 ± 1.8°	0.08	>10	>10	2.4°

Φ: index of dependability; AE: absolute error; CE: constant error; LS: lumbar spine; MCI: Movement control impairment; MDC: minimal detectable change; RE: Reposition Error; RMSJ: root mean squared jerk; ROM: range of motion; SD: Standard deviation; TS: Thoracic Spine.

further for intersubject comparisons and monitoring changes after an intervention.

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Appendix A. Supplementary data

Supplementary data related to this article can be found at http://dx.doi.org/10.1016/j.math.2016.02.013.

References

Airaksinen O, Brox JI, Cedraschi C, Hildebrandt J, Klaber-Moffett J, Kovacs F, et al. European guidelines for the management of chronic nonspecific low back pain. Eur Spine J 2006;15(Suppl. 2):S192–300 [Chapter 4].
Al Zoubi FM, Preuss RA. Reliability of a measure of total lumbar spine range of motion in individuals with low back pain. J Appl Biomech 2013;29:670–7.
Bauer CM, Rast FM, Ernst MJ, Kool J, Oetiker S, Rissanen SM, et al. Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement. J Electromyogr Kinesiol 2015;25:782–90.
Brennan R. Generalizability theory. New York: Springer-Verlag; 2001.
Carter RE, Lubinsky J, Domholdt E. Rehabilitation research: principles and applications. 4th ed. St. Louis, MO: Elsevier Saunders; 2005.
Costa Lda C, Maher CG, McAuley JH, Hancock MJ, Herbert RD, Refshauge KM, et al. Prognosis for patients with chronic low back pain: inception cohort study. BMJ 2009;339:b3829.
Crawford NR, Yamaguchi GT, Dickman CA. A new technique for determining 3-D joint angles: the tilt/twist method. Clin Biomech (Bristol, Avon) 1999;14:153–65.
de Vet HC, Terwee CB, Knol DL, Bouter LM. When to use agreement versus reliability measures. J Clin Epidemiol 2006;59:1033–9.
Ernst M, Rast F, Bauer C, Marcar V, Kool J. Determination of thoracic and lumbar spinal processes by their percentage position between C7 and the PSIS level. BMC Res Notes 2013;6:58.
Koumantakis GA, Winstanley J, Oldham JA. Thoracolumbar proprioception in individuals with and without low back pain: intratester reliability, clinical applicability, and validity. J Orthop Sports Phys Ther 2002;32:327–35.
Laird RA, Gilbert J, Kent P, Keating JL. Comparing lumbo-pelvic kinematics in people with and without back pain: a systematic review and meta-analysis. BMC Musculoskelet Disord 2014;15:229.
Littlewood C, May S. Measurement of range of movement in the lumbar spine-what methods are valid? A systematic review. Physiotherapy 2007;93:201–11.
Luomajoki H, Kool J, de Bruin E, Airaksinen O. Reliability of movement control tests in the lumbar spine. BMC Musculoskelet Disord 2007;8:90.
Madgwick S, Vaidyanathan R, Harrison A. An efficient orientation filter for IMU and MARG sensor arrays. Department of Mechanical Engineering, University of Bristol; 2010.
May S, Littlewood C, Bishop A. Reliability of procedures used in the physical examination of non-specific low back pain: a systematic review. Aust J Physiother 2006;52:91–102.
O'Sullivan P. Diagnosis and classification of chronic low back pain disorders: maladaptive movement and motor control impairments as underlying mechanism. Man Ther 2005;10:242–55.
Oesch P, Hilfiker R, Keller S, Kool J, Schädler S, Tal-Akabi A, et al. Assessments in der muskuloskelettalen rehabilitation. 1st ed. Verlag Hans Huber; 2007.
Rausch Osthoff AK, Ernst MJ, Rast FM, Mauz D, Graf ES, Kool J, et al. Measuring lumbar reposition accuracy in patients with unspecific low back pain: systematic review and meta-analysis. Spine (Phila Pa 1976) 2015;40:E97–111.
Roetenberg D, Baten CT, Veltink PH. Estimating body segment orientation by applying inertial and magnetic sensing near ferromagnetic materials. IEEE Trans Neural Syst Rehabil Eng 2007;15:469–71.
Sahrmann S. Diagnosis and treatment of movement impairment syndromes. St. Louis, London, Philadelphia, Sydney, Toronto: Mosby; 2002.
Schelldorfer S, Ernst MJ, Rast FM, Bauer CM, Meichtry A, Kool J. Low back pain and postural control, effects of task difficulty on centre of pressure and spinal kinematics. Gait Posture 2015;41:112–8.
Seffinger MA, Najm WI, Mishra SI, Adams A, Dickerson VM, Murphy LS, et al. Reliability of spinal palpation for diagnosis of back and neck pain: a systematic review of the literature. Spine 2004;29:413–25.
Slaboda JC, Boston JR, Rudy TE, Lieber SJ, Rasetshwane DM. The use of splines to calculate jerk for a lifting task involving chronic lower back pain patients. IEEE Trans Neural Syst Rehabil Eng 2005;13:406–14.
Stochkendahl MJ, Christensen HW, Hartvigsen J, Vach W, Haas M, Hestbaek L, et al. Manual examination of the spine: a systematic critical literature review of reproducibility. J Manip Physiol Ther 2006;29:475.
van Trijffel E, Anderegg Q, Bossuyt PM, Lucas C. Inter-examiner reliability of passive assessment of intervertebral motion in the cervical and lumbar spine: a systematic review. Man Ther 2005;10:256–69.
Vibe Fersum K, O'Sullivan PB, Kvale A, Skouen JS. Inter-examiner reliability of a classification system for patients with non-specific low back pain. Man Ther 2009;14:555–61.

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Pain intensity attenuates movement control of the lumbar spine in low back pain

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ABSTRACT

Introduction: Pain intensity attenuates muscular activity, proprioception, and tactile acuity, with consequent changes of joint kinematics. People suffering from low back pain (LBP) frequently show movement control impairments of the lumbar spine in sagittal plane. This cross-sectional, observational study investigated if the intensity of LBP attenuates lumbar movement control. The hypothesis was that lumbar movement control becomes more limited with increased pain intensity. *Methods*: The effect of LBP intensity, measured with a numeric rating scale (NRS), on lumbar movement control was tested using three movement control tests. The lumbar range of motion (ROM), the ratio of lumbar and hip ROM as indicators of direction specific movement control, and the recurrence and determinism of repetitive lumbar movement patterns were assessed in ninety-four persons suffering from LBP of different intensity and measured with an inertial measurement unit system. Generalized linear models were fitted for each outcome. *Results*: Lumbar ROM (+0.03°, $p = 0.24$) and ratio of lumbar and hip ROM (0.01, $p = 0.84$) were unaffected by LBP intensity. Each one point increase on the NRS resulted in a decrease of recurrence and determinism of lumbar movement patterns (−3.11 to −0.06, $p \leqslant 0.05$). *Discussion*: Our results indicate changes in movement control in people suffering from LBP. Whether decreased recurrence and determinism of lumbar movement patterns are intensifiers of LBP intensity or a consequence thereof should be addressed in a future prospective study.

1. Introduction

Low back pain (LBP) is a common disorder with a lifetime prevalence as high as 84%, and a high probability of recurrence (Airaksinen et al., 2006). In many cases the cause of pain is never fully resolved (Hoy et al., 2010). LBP causes functional impairment in everyday life for a large proportion of the population and thus imposes large demands on healthcare and social systems (Dunn and Croft, 2004). Contemporary LBP classification systems propose that there is a large group of patients who present with movement control impairments (MCI), which are a relevant and provocative factor for ongoing pain (O'Sullivan, 2005). Typically 50% of patients with a MCI demonstrate changes in the sagittal plane (Vibe Fersum et al., 2009). These impairments may be the consequence of decreased tactile acuity (Luomajoki and Moseley, 2011), decreased ability to modulate task specific proprioceptive feedback (Claeys et al., 2011) or altered muscle recruitment patterns (Humphrey et al., 2005).

Tests of direction specific movement control (DSMC) assess the ability of a person to stabilize the lumbar spine during active movement of the hip and or knee. They are based on visual observation and use a dichotomous rating, have substantial reliability, and have been shown to differentiate between asymptomatic

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persons and patients with LBP (Luomajoki et al., 2007, 2008). However, objective, quantitative data on the severity of MCI assessed by DSMC tests in people suffering from LBP are currently lacking. Repetitive movements (RM) can demonstrate changes in lumbar spine kinematics which are not observed when analyzing purely the range of motion or magnitude of MCI (Lamoth et al., 2006; Silfies et al., 2009). Less variable movement patterns of lumbar spine were observed in persons with chronic LBP when they repetitively picked up a box (Dideriksen et al., 2014) or performed repeated trunk movements (Asgari et al., 2015). Persons with chronic LBP also demonstrated less variable recruitment patterns of lumbar erector muscles during lifting tasks (Falla et al., 2014).

The effect of LBP on lumbar movement may be more pronounced in higher order kinematics (Aluko et al., 2013; Bourigua et al., 2014; Marras et al., 1993, 1995). Participants with chronic LBP showed smaller lumbar angular velocity and acceleration during a repeated trunk flexion–extension task, compared to pain free participants. These group differences were less pronounced when analyzing purely their angular displacement (Marras et al., 1995). Increased lumbar angular velocity and acceleration during lifting tasks had a greater odds ratio for future low back pain episodes when compared to changes in angular displacement (Marras et al., 1993). Chronic LBP patients showed lower angular velocity during trunk flexion at self-selected and fast movement speeds (Bourigua et al., 2014). Lumbar acceleration increased after a six weeks exercise intervention that reduced LBP intensity (Aluko et al., 2013).

Previous cross-sectional studies often do not report the relationship between LBP intensity and MCI, and do not consider that pain differently attenuates motor planning and diminishes proprioception, and that tactile acuity depends up on its intensity (Catley et al., 2014; Matre et al., 2002; Ervilha et al., 2004). The purpose of this study is to investigate the effect of LBP intensity on MCI using two DSMC tests, and one RM test. The emphasis is on reduced control of active movement (Luomajoki et al., 2008; O'Sullivan, 2005) and on repetitive task movement control (Dideriksen et al., 2014). It is hypothesized that lumbar movement control deteriorates with increased LBP intensity. Anthropometric factors such as age, gender, or body mass index (BMI) influence lumbar kinematics (Consmuller et al., 2012). Persons engaging in heavy manual labor have a higher risk of developing LBP (Hoozemans et al., 2002). These factors should be controlled for when investigating the relationship between lumbar kinematics and LBP.

2. Methods

2.1. Design

Cross-sectional, observational study.

2.2. Participants

Sixty-three participants with sub-acute or chronic LBP and 31 asymptomatic participants, aged between 18 and 65 years were recruited from physiotherapy practice, the university campus and through newspaper advertisements. Participants with LBP were included if their current episode of LBP persisted for four weeks or longer, and if they reported at least moderate disability, defined as an Oswestry-disability-index (ODI) >8% and a low level of psychosocial risk factors defined with less than four points on the subscale of the STarT Back screening tool (Mannion et al., 2006). Exclusion criteria were specific LBP, vertigo or disturbance of the equilibrium, systemic diseases (diabetes, tumours), pain in other areas of the body (neck, head, thoracic spine, or arms), complaints, injury, or surgery of the legs (hips to feet) within the last six months, medication affecting postural control (e.g. antidepressants) and pregnancy. The exclusion criteria for asymptomatic participants were the same as for the LBP participants, and additionally no current LBP episodes or episodes during the preceding three months. The study was conducted according to the declaration of Helsinki, and approved by the local ethics committee (KEK-ZH-2011-0522). Participants provided their written informed consent.

2.3. Movement analysis

2.3.1. Sensor placement and data processing

Trunk movements were measured by an inertial measurement unit (IMU) system, with multiple IMUs placed above the right thigh, sacrum and at the level of L1, (Ernst et al., 2013; Schelldorfer et al., 2015) (Fig. 1). The IMU system has been shown to provide concurrently valid estimates of spinal kinematics (Bauer et al., 2015).

The sensors of the IMU system (ValedoMotion, Hocoma AG, Volketswil, Switzerland) include a tri-axial gyroscope, magnetometer, and accelerometer. Movement data were recorded with a sampling frequency of 200 Hz (Valedo®Research, Hocoma AG). The raw data from the IMUs were transformed into quaternions to prevent rotational singularities (Madgwick et al., 2010). Segmental kinematics were calculated using the tilt/twist formulation (Crawford et al., 1999) with sagittal and frontal planes defined by the global coordinate system. All outcome variables were derived from the flexion/extension angle, where flexion is positive and extension is negative. An angle of zero degrees is defined as

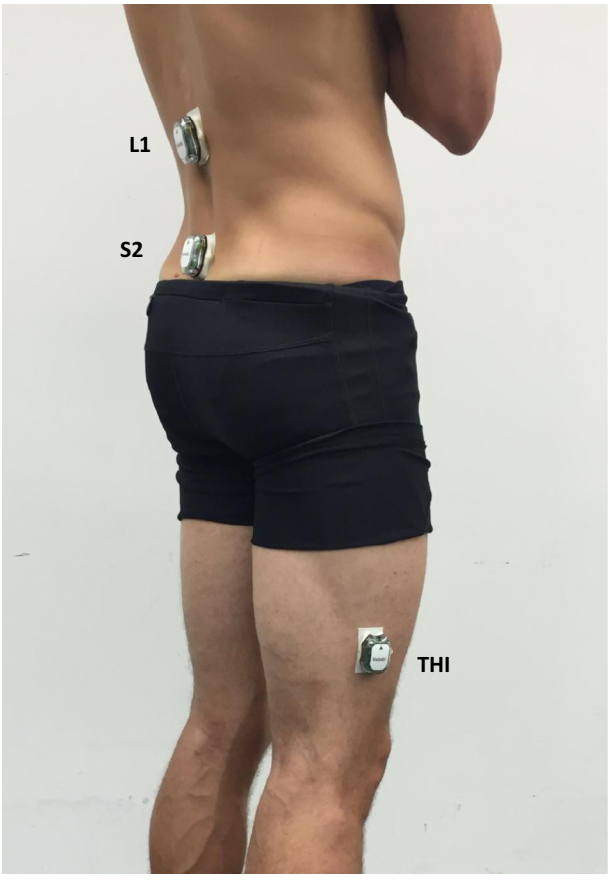

Fig. 1. Experimental setup: IMUs were placed on the right thigh (THI), and level of sacrum (S2), and L1 (L1).

alignment of two IMUs. A second-order zero-phase low-pass Butterworth filter (1 Hz cut-off frequency) was applied on the angular displacement data since angular velocity and acceleration required smoothing to obtain interpretable estimates. Angular velocity and acceleration were calculated using the first and second derivative of the filtered angular displacement data. A complete description of the data processing from raw data to tilt/twist angles is described elsewhere (Bauer et al., 2015).

2.3.2. Movement tests

Participants attended one measurement session and performed two DSMC tests: "Sitting Knee Extension", and "Waiters Bow"; and one RM test: "Pick Up a Box" (Fig. 2) (Bauer et al., 2015). Prior to each test the participants received standardized oral instructions by one of the examiners and visual instructions in a video. In case of poor initial performance these instructions were repeated up to three times and the test was demonstrated by one examiner. If the participant was still performing the test incorrectly it was omitted.

During "Sitting Knee Extension" the participants sat upright and were asked to stabilize their lumbar spine whilst extending their right knee. They were instructed to stop extending their knee before they perceived movement of their lumbar spine, without giving a target range for knee extension or target duration for the test. In "Waiters Bow" they were instructed to stand upright and then flex their hips as far as possible whilst keeping their lumbar spine stable. They were instructed to stop flexing their hips before they perceived movement of their lumbar spine, without giving a target range for hip flexion or target duration for the test. The participants were allowed to perform the DSMC tests at their own preferred speed.

The "Pick Up a Box" test consisted of ten cycles, of four seconds duration, starting in upright standing. During each cycle the participants were asked to pick up the box from the ground and put it back down again. They were guided with a metronome set at 60 bpm. The box was loaded to ten percent of their body weight and placed at a standardized distance in front of the participants.

The order of the tests was randomized between participants. The DSMC tests were repeated three times and the participants were allowed to choose their rest time between repetitions, whereas the RM test was performed one time.

2.3.3. Outcomes

For "Sitting Knee Extension" the range of motion at the lumbar spine $ROM\,LS$ was calculated between the sacrum and L1 sensors. For "Waiters Bow" the ratio between the range of motion of the lumbar spine and the hip was calculated and is later denoted with $\frac{LS}{Hip}ROM$. For both outcomes, the mean of the three repetitions was used for further analysis. For the "Pick Up a Box" test recurrence quantification analysis was performed on the angular displacement, velocity, and acceleration data. This method has been described previously and is only briefly summarized here (Webber and Zbilut, 1994). In recurrence quantification analysis, movement data are projected into a phase space by taking time-delayed samples from the movement data. The time-delayed samples represent movement patterns which can be visualized as points in the phase-space plot. Similar movement patterns are located close to each other, and form a cluster of recurrent points ($R_{i,j}$). In this study, the phase-space reconstruction was undertaken separately for angular displacement, velocity, and acceleration data by using the set of parameters specified in Table 1. All $R_{i,j}$:s were subsequently transferred into a $N \times N$-sized recurrence plot (RP) with N being the number of measurement points. Two measures were then calculated: the recurrence rate (REC) and the determinism (DET). REC is a measure of the density of the recurrent points in the RP. It measures the probability of recurrence of movement patterns and is expressed as:

$$REC = \frac{1}{N^2}\sum_{i,j=1}^{N} R_{i,j} * 10^2$$

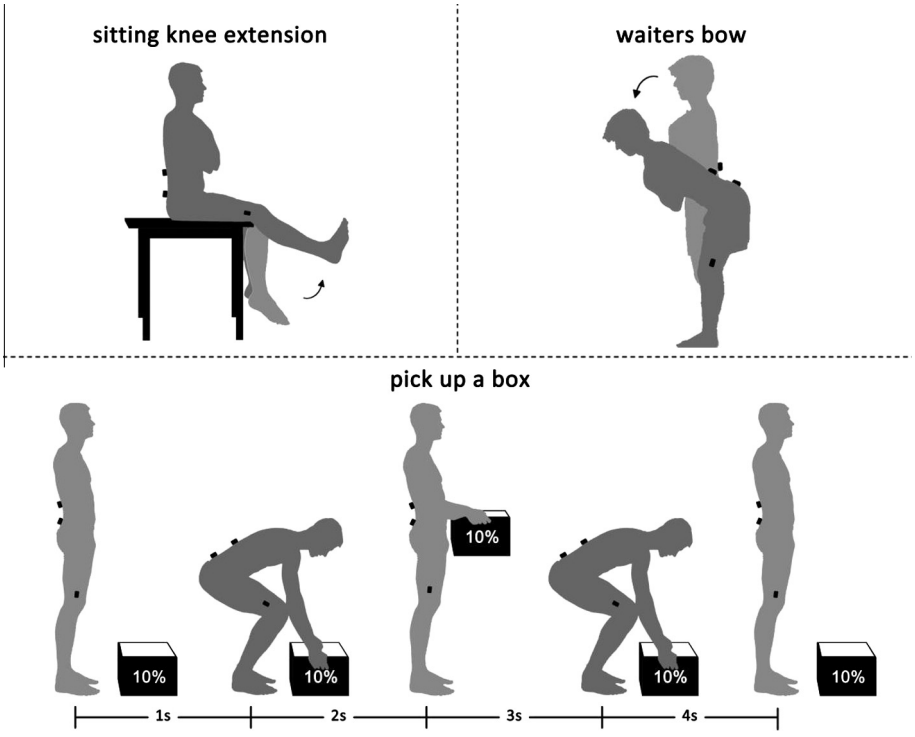

Fig. 2. Test procedure "Sitting Knee Extension", "Waiters Bow" and "Pick Up a Box".

Table 1
Input parameters used in recurrence quantification analysis.

Picking up a box	Delay	Embedding dimension	Distance	*lmin*	Size of neighborhood
Angular displacement	37	2	Euclidian	20	1.3
Angular velocity	16	2	Euclidian	50	1.3
Angular acceleration	9	2	Euclidian	20	1.3

lmin – minimal length of diagonal line.
The delays were estimated using mutual information analysis. The first minimum of mutual information was defined as the optimal delay. The embedding dimensions were estimated by calculating the correlation dimension with different embedding dimensions. The optimal value of embedding dimension was chosen as the starting point where the correlation dimension did not increase significantly although increasing the embedding dimension.

DET is the amount of $R_{i,j}$ that form diagonal lines (i.e. are sequential to each other in time) of a prespecified minimal length (*lmin*) given in Table 1. The DET is a measure of the stochasticity of the movement data and expressed as:

$$DET = \frac{\sum_{l=lmin}^{lmax} l * P(l)}{\sum_{l=1}^{lmax} l * P(l)} * 10^2$$

with l being the length of the diagonal lines, *lmax* the maximal possible length of the diagonal lines, and $P(l)$ being the number of diagonal lines of length l. All data processing and calculations were done using Matlab 2012b® (Mathworks, USA), with code from University of Potsdam, Germany (Marwan and Kurths, 2002). *REC* and *DET* were calculated for angular displacement (REC AD and DET AD), velocity (REC AV and DET AV), and acceleration (REC AA and DET AA). In a previous study the reliability of all outcomes, using the current test setup, was found to be high (Bauer et al., 2015).

2.4. Covariates

For each of the participants, LBP intensity, age, gender, BMI, and the amount of physical stress at work (PS) were recorded. All participants rated their LBP intensity, defined as the mean level of LBP pain during the past four weeks, using a 11 point numerical rating scale (NRS) anchored with "no pain" (0) through to "the worst possible pain imaginable" (10). Following this the participants were allocated into eight groups, according to their perceived LBP intensity (0–7). PS was measured with a five point Likert scale; ranging from "almost no physical stress" (1) to "maximal physical stress" (5) (Galati-Petrecca, 2008).

2.5. Statistical analysis

For each outcome a linear model was fitted to the data with LBP intensity as the covariate of interest. In a first model, we adjusted for gender, age, BMI, PS and all the two-way interactions between these covariates with LBP intensity. A stepwise model selection procedure with backwards optimization by the Akaike-Information-criterion was used to determine the final model. The aim of this procedure was to choose a parsimonious model in order to prevent overfitting of the data. This procedure ensured that the model is optimized for prediction, that is, for future data.

Therefore the model for each observation of the outcome Y_i was

$$\begin{aligned} Y_i = {} & \beta_0 + \beta_1 \mathrm{NRS}_i + \beta_2 Gender_i + \beta_3 Age_i + \beta_4 \mathrm{BMI}_i + \beta_5 \mathrm{PS}_i \\ & + \beta_6 \mathrm{NRS} \times Gender_i + \beta_7 \mathrm{NSR} \times Age_i + \beta_8 \mathrm{NRS} \times \mathrm{BMI}_i \\ & + \beta_9 \mathrm{NRS} \times \mathrm{PS}_i + \varepsilon_i \end{aligned}$$

with β_0 representing the intercept, β_k the effect of the k-covariate and ε_i the independent and normal distributed errors $\varepsilon_i \sim N(0, \sigma^2)$.

Residual analysis was performed to check the models assumptions. Therefore the log-transformed ratio $\frac{LS}{Hip} ROM$ was modeled since the residuals did not have a normal distribution. Point and interval estimations were performed for each covariate. The alpha-level was set at 0.05. Statistical analysis was done using R (R Foundation for statistical computing, Austria).

3. Results

Sixty three persons with LBP and thirty one pain-free persons were included. The distribution of LBP intensity and the descriptive data of the covariates and outcomes are shown in Table 2. Fig. 3 depicts the DSMC and RM angular displacement, velocity, and acceleration trajectories of one representative participant with high movement control and one participant with low movement control. The parameter estimates for the final model for each outcome are shown in Table 3. Depending upon the presence of interaction terms, the observed effect of a one point increase of LBP intensity ($\hat{\omega}$) can be a function of age, BMI and gender, but not of PS (Table 3).

Sitting Knee Extension:

$\hat{\omega}$ for *ROM LS* was

$$\hat{\omega}_{ROM\,LS} = 0.3^\circ - 0.5^\circ * 1_{Gender=Female}$$

with 1 being the indicator function, indicating a 0.3° increase in males, but a −0.2° decrease in females. This means that LBP intensity had no significant effect on *ROM LS*.

Waiters Bow:

$\hat{\omega}$ for $\frac{LS}{Hip} ROM$ was

$$\hat{\omega}_{\frac{LS}{Hip}ROM} = 0.0 + 0.1 * 1_{Gender=Female}$$

This indicates a 0.1 increase for females on the log scale and no changes for males. This means that LBP intensity had no significant effect on $\frac{LS}{Hip} ROM$.

Pick Up a Box:

$\hat{\omega}$ for *REC AD* and *DET AD* were

$$\hat{\omega}_{REC\,AD} = 0.11$$

$$\hat{\omega}_{DET\,AD} = -0.06$$

Due to the absence of interactions, these effects were independent of age, BMI and gender. This means that *DET AD* significantly decreased with increasing LBP intensity ($p = 0.01$).

$\hat{\omega}$ for *REC AV and DET AV* were

$$\hat{\omega}_{REC\,AV} = -0.25$$

$$\hat{\omega}_{DET\,AV} = -3.11 + 0.11 * \mathrm{BMI}$$

$\hat{\omega}_{REC\,AV}$ was independent of age, gender or BMI. $\hat{\omega}_{DET\,AV}$ was dependent on BMI. For example, the effect of a one point increase in LBP intensity for a person with a BMI of 19.0 is −1.02 while for a person with a BMI of 23 is −0.58. The main effect of LBP were statistically significant ($p = 0.03$ and $p = 0.05$) while the interaction of LBP with BMI was not.

$\hat{\omega}$ for *REC AA* was

$$\hat{\omega}_{REC\,AA} = -2.03 + 1.14 * 1_{Gender=Female} + 0.03 * Age$$

and thus a function of age and gender. For example, the effect of a one point increase in LBP intensity for a 20 year old female is −0.29; while for a 50 year old female it is 0.61. The main effect of LBP and the interaction of LBP with gender were statistically significant

Table 2
Descriptive statistics.

LBP	*n*	*Gender*	*Age*	BMI	PS	Sitting knee extension	Waiters bow	Picking up a box					
NRS		m/f	years	(kg/m^2)		ROM LS (°)	Ratio LS/hip ROM	*REC* AD	*DET* AD	*REC* AV	*DET* AV	*REC* AA	*DET* AA
0	31	14/17	40.1(±12.1)	22.7(±2.9)	1(1–4)	2.6(±3.7)	0.3(±0.2)	41.0(±1.3)	98.7(±0.4)	43.5(±2.0)	94.0(±3.5)	39.2(±5.1)	58.3(±9.3)
1	4	3/1	49.8(±10.6)	26.9(±2.8)	1(1–2)	1.2(±3.8)	0.5(±0.4)	40.1(±0.4)	98.4(±0.4)	42.2(±3.4)	87.1(±9.6)	39.8(±5.9)	55.4(±11.8)
2	19	11/8	43.8(±13.2)	24.9(±3.7)	1(1–3)	2.2(±3.0)	0.2(±0.1)	41.7(±2.3)	98.1(±0.5)	43.7(±2.5)	89.6(±6.0)	37.4(±1.9)	52.4(±6.4)
3	13	7/6	35.2(±10.8)	24.1(±4.8)	1(1–4)	2.5(±2.7)	0.5(±0.5)	41.4(±1.3)	98.3(±0.4)	42.9(±1.9)	93.2(±3.7)	37.9(±2.6)	54.4(±6.3)
4	15	7/8	34.6(±11.0)	22.8(±3.2)	2(1–4)	3.3(±4.0)	0.3(±0.2)	41.9(±1.3)	98.4(±0.3)	42.5(±1.9)	90.6(±3.7)	38.5(±2.6)	52.8(±7.6)
5	5	2/3	38.0(±15.8)	23.8(±2.9)	1(1–2)	1.1(±2.7)	0.8(±0.7)	40.9(±0.6)	98.3(±0.2)	41.5(±1.9)	89.4(±4.0)	38.7(±2.8)	53.8(±5.5)
6	4	1/3	45.0(±11.6)	26.5(±7.3)	2(1–5)	2.4(±2.7)	0.7(±0.7)	41.4(±1.4)	98.2(±0.3)	42.6(±1.7)	90.4(±3.4)	38.7(±4.4)	51.2(±9.3)
7	3	1/2	30.0(±4.0)	21.9(±1.2)	2(2–5)	1.3(±3.5)	0.4(±0.2)	41.9(±1.6)	98.1(±0.3)	41.6(±1.3)	86.7(±6.9)	35.7(±1.4)	49.9(±4.8)

AA – angular acceleration; AD – angular displacement; AV – angular velocity; BMI – body mass index; DET – determinism; LBP – mean low back pain in the past four weeks; LS – lumbar spine; NRS – numeric pain rating scale; PS – physical stress at work; REC – recurrence rate; ROM – range of motion.
Results are provided as median (range) or mean (±standard deviation).

($p = 0.03$ and $p = 0.02$), while the interaction of LBP with age was not. This means that *REC AA* either increased or decreased with increasing LBP intensity, depending on the age of the participant.

$\hat{\omega}$ for *DET AA* was

$$\hat{\omega}_{DET\,AA} = -0.86$$

This means that *DET AD* significantly decreased with increasing LBP intensity ($p = 0.05$).

In summary the results show a statistically significant effect of LBP intensity on *REC* and *DET*. REC AV and *DET* decrease with increasing LBP intensity whilst *REC AA* either increases or decreases, depending on the age of the participant.

4. Discussion

This study examined if the intensity of LBP affects movement control of the lumbar spine during two DSMC tests and one RM test. LBP intensity had no effect on DSMC, which is unexpected since previous research demonstrated reduced DSMC in patients with LBP (Luomajoki et al., 2008). This can be explained by methodological differences regarding the group allocation, study population, and measurement systems. In our study, participants were allocated into eight groups according to their LBP intensity, while Luomajoki and colleagues (2008) summarized chronic LBP patients with varying degrees of pain intensity into one group, which hampers comparability between the results. Dichotomization of the participants might have increased the contrast between the two groups. However, using a quantitative approach leads to a more detailed insight into the relation between MCI and LBP.

Observed group differences might be further increased by the selection of the study subjects. Luomajoki and colleagues (2008) recruited LBP patients that were referred from physicians and treated by physiotherapists. Conversely, not all participants with LBP recruited for the present study perceived their condition serious enough to seek treatment, indicating a lower burden of disease, with less impairment due to LBP. Luomajoki and colleagues (2008) used a dichotomous rating of movement control by observation of lumbar spine flexion, while in the present study an IMU system was used and movement control was measured continuously.

These findings raise the possibility that (i) a relation between severity of DSMC impairment and LBP intensity exists, (ii) DSMC is a clinically relevant feature, but (iii) DSMC impairment becomes clinically relevant only after it exceeds a certain magnitude or cut-off point. It is possible that only a smaller subgroup of patients with LBP show a DSMC impairment that manifests in "Waiters Bow" and "Sitting Knee Extension". The link between performance in DSMC tests and LBP intensity is based upon investigations of tactile acuity tests (Catley et al., 2014; Luomajoki and Moseley, 2011). However our results do not validate this link.

Recurrence and determinism of lumbar movement patterns were significantly affected by LBP intensity. More variable lumbar movement patterns, indicated by reduced recurrence and determinism, were found with increasing levels of LBP and this effect was more pronounced in angular velocity and acceleration data. Silfies et al. (2009) found that the variability of lumbar movement during a repetitive reaching task was increased in LBP patients, when compared to those without pain indicating impaired movement control (Silfies et al., 2009). Lamoth and colleagues (2006) revealed that lumbar angular velocity patterns during gait were more variable in LBP patients, compared to no pain, and found these changes to be related to poor coordination of lumbar erector spinae muscles (Lamoth et al., 2006). These findings are in line with the results of the present study, although both studies used a different methodology, regarding task and calculation of variability of lumbar movement.

Falla and colleagues (2014) examined lumbar muscular activation patterns during a RM test and found less variability in LBP. To maintain constant movement patterns during repetitive activities an adaptive muscular activation is necessary, leading to a greater variability of muscular activity and less variability of movement patterns. Thus, stereotyped movement may be accompanied by variable electromyography patterns, while this may be reversed in painful conditions such as LBP (Falla et al., 2014).

Contrary to our findings one study reported less variability using a similar RM test (Dideriksen et al., 2014). Differences in the data processing may explain this contradiction, as Dideriksen and colleagues used a notch filter to smooth out the frequencies related to the RM, without affecting other frequency components. In this way only the deviation from the target movement was investigated (Dideriksen et al., 2014), while the present study investigated the target movement.

Participants in the present study were asked to perform the repetitive movement with a predetermined, fast speed, possibly overriding any protective feedforward strategy of movement control. This contrasts findings from two studies (Arzi et al., 2014; Uri et al., 2015) who showed that patients after shoulder surgery had less variable kinematics of the shoulder joint when moving at self-selected slow and fast speed. To test this hypothesis the repetitive "Pick up a box" task should be performed at self-selected preferred, slow, and fast speed.

In summary the results indicate that there is an effect of perceived intensity of LBP on lumbar movement control. This effect manifests in the variability of lumbar movement patterns, but not in DSMC. RM tests, in contrast to DSMC tests, might better reflect lumbar movement control in activities of daily living, which in turn might be of greater relevance in the development, persistence and intensity of LBP (O'Sullivan, 2005).

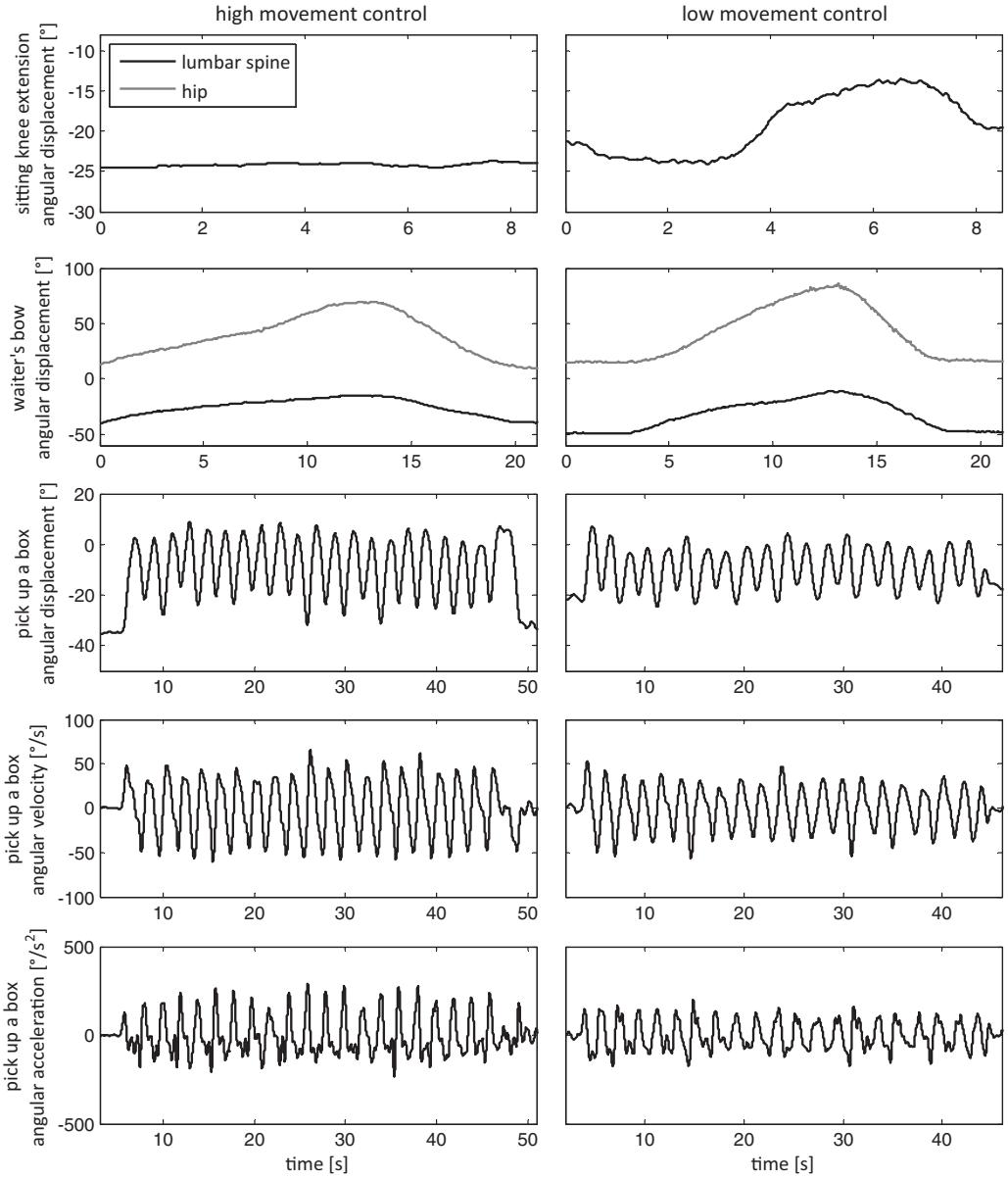

Fig. 3. Comparison of movement control. The left column shows a participant without low back pain and high movement control, the right column a participant with high intensity low back pain and low movement control.

The final models show that covariates such as gender and BMI, significantly affect movement control. Their effect was not consistent across all measures of movement control, and sometimes exceeded that of LBP intensity. Consequently it is recommended to consider these covariates in future research on movement control. Other covariates, such as the frequency and duration of the current LBP episode, physical stress during leisure time, might also be related to MCI. Furthermore anthropometric factors, such as a participants arm length, might impact performance during a repeated lifting test and should be controlled for in future research. The models and subsequent interpretations were based on the assumption of a linear relationship between perceived LBP intensity and MCI, which was confirmed by partial-residual plots. Backwards selection of covariates enabled us to test the effect of two-way interactions before testing main effects, and to exclude redundant covariates from the final models. Perceived LBP intensity was measured using a NRS and may not have ratio qualities (Price et al., 1994). Therefore a one point increase in mild pain intensity may not have the same meaning as in high pain intensity. In addition, two participants with similar pain might not rate their pain equally.

The number of participants was unevenly distributed across the levels of perceived LBP intensity, with a small number that rated their perceived LBP higher than five. However, the distribution of an outcome does not affect the models validity, provided that the residuals follow a normal distribution, verified by residual analysis. Exclusion criteria were asserted using patient history interviews and questionnaires. To improve validity of patient selection, ascertainment should be accompanied by anamnestic interviews, physical examination, imaging techniques or other instruments.

This study investigated sagittal plane MCI as this was found to be an important subgroup of MCI (Vibe Fersum et al., 2009). Future studies should expand on this research and address control of combined movements since LBP and injury might occur while combining rotational torques and sagittal or lateral rotations. The IMU system provides valid and reliable estimates of lumbar movement control in the sagittal plane (Bauer et al., 2015), while validity and reliability of combined movements have not been addressed until

Table 3
Final model of each outcome.

Test & variable	Covariate	Point estimation	95% CI LB	95% CI UB	*p*-Value
Sitting knee extension					
ROM LS (°)	LBP	0.3	−0.2	0.9	0.24
	Gender (Female)	0.8	−1.3	2.9	0.44
	BMI	−0.2	−0.4	0.0	0.05*
	LBP:Gender (Female)	−0.5	−1.3	0.2	0.14
Waiters bow					
Log ratio LS/Hip					
ROM	LBP	0.0	−0.1	0.2	0.84
	Gender (Female)	−0.1	−0.6	0.5	0.78
	BMI	0.1	0.0	0.1	0.03*
	PS	−0.4	−0.7	0.0	0.04*
	LBP:Gender (Female)	0.1	−0.1	0.3	0.16
Pick up the box					
REC AD	LBP	0.11	−0.05	0.26	0.18
DET AD	LBP	−0.06	−0.11	−0.02	0.01*
	Age	−0.01	−0.01	0.00	0.11
REC AV	LBP	−0.25	−0.46	−0.03	0.03*
DET AV	LBP	−3.31	−6.21	−0.01	0.05*
	Gender (Female)	−2.22	−4.25	−0.02	0.03*
	BMI	−0.66	−1.11	−0.25	0.45
	NRS:BMI	0.11	−0.02	0.24	0.11
REC AA	LBP	−2.03	−3.87	−0.20	0.03*
	Gender (Female)	−2.35	−5.10	0.47	0.09
	Age	−0.05	−0.16	0.06	0.40
	LBP:Gender (Female)	1.14	0.17	2.12	0.02*
	LBP:Age	0.03	−0.01	0.07	0.14
DET AA	LBP	−0.86	−1.73	0.00	0.05*
	BMI	−0.59	−1.06	−0.13	0.01*

Abbreviations: 95% CI – 95% confidence interval; AA – angular acceleration; AD – angular displacement; AV – angular velocity; BMI – body mass index; DET – determinism ; LB – lower bound; LBP – low back pain intensity; LS – lumbar spine; PS – physical stress at work; REC – recurrence rate; ROM – range of motion; UB – upper bound.
The reference level for gender was defined as female. The point estimation for each covariate is the effect of a one point increase of the respective covariate on the variable. For gender it represents the effect of being female.
* $p \leqslant 0.05$.

now. Choosing an appropriate filtering technique is a compromise between loss of information and noise allowed through. While significant associations between LBP intensity and MCI were found following our procedure, we might have missed small fragmentations of movement related to LBP (Dideriksen et al., 2014). Future studies should address options that might conserve such information.

5. Conclusion

The effect of perceived LBP intensity on lumbar movement control was analyzed and controlled for the effect of age, gender, BMI and PS. A linear effect of LBP intensity on variability of lumbar movement patterns was found, but not on DSMC. The variability of lumbar movement patterns increased with greater LBP intensity, measured with a repetitive Pick up the box test and recurrence quantification analysis.

Conflict of interest

The authors declare that they have no conflict of interest.

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References

Airaksinen O, Brox JL, Cedraschi C, Hildebrandt J, Klaber-Moffett J, Kovacs F, et al. European guidelines for the management of chronic nonspecific low back pain. Eur Spine J 2006;15Suppl(2):S192–300 [Chapter 4].
Aluko A, DeSouza L, Peacock J. The effect of core stability exercises on variations in acceleration of trunk movement, pain, and disability during an episode of acute nonspecific low back pain: a pilot clinical trial. J Manipulative Physiol Ther 2013;36(8):497–504 [e491-493].
Arzi H, Krasovsky T, Pritsch M, Liebermann DG. Movement control in patients with shoulder instability: a comparison between patients after open surgery and nonoperated patients. J Shoulder Elbow Surg 2014;23(7):982–92.
Asgari M, Sanjari MA, Mokhtarinia HR, Moeini Sedeh S, Khalaf K, Parnianpour M. The effects of movement speed on kinematic variability and dynamic stability of the trunk in healthy individuals and low back pain patients. Clin Biomech 2015;30(7):682–8.
Bauer CM, Rast FM, Ernst MJ, Kool J, Oetiker S, Rissanen S, et al. Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement. J Electromyogr Kinesiol 2015;25(5):782–90.
Bourigua I, Simoneau EM, Leteneur S, Gillet C, Ido G, Barbier F. Chronic low back pain sufferers exhibit freezing-like behaviors when asked to move their trunk as fast as possible. Spine J 2014;14(7):1291–9.
Catley M, O'Connell N, Berryman C, Ayhan FF, Moseley GL. Is tactile acuity altered in people with chronic pain? A systematic review and meta-analysis. J Pain 2014;15(10):985–1000.
Claeys K, Brumagne S, Dankaerts W, Kiers H, Janssens L. Decreased variability in postural control strategies in young people with non-specific low back pain is associated with altered proprioceptive reweighting. Eur J Appl Physiol 2011;111(1):115–23.
Consmuller T, Rohlmann A, Weinland D, Druschel C, Duda GN, Taylor WR. Velocity of lordosis angle during spinal flexion and extension. PLoS ONE 2012;7(11):e50135.
Crawford NR, Yamaguchi GT, Dickman CA. A new technique for determining 3-D joint angles: the tilt/twist method. Clin Biomech 1999;14(3):153–65.
Dideriksen JL, Gizzi L, Petzke F, Falla D. Deterministic accessory spinal movement in functional tasks characterizes individuals with low back pain. Clin Neurophysiol 2014;125(8):1663–8.
Dunn KM, Croft PR. Epidemiology and natural history of low back pain. Eura Medicophys 2004;40(1):9–13.
Ernst MJ, Rast FM, Bauer CM, Marcar VL, Kool J. Determination of thoracic and lumbar spinal processes by their percentage position between C7 and the PSIS level. BMC Res Notes 2013;6:58.

Ervilha UF, Arendt-Nielsen L, Duarte M, Graven-Nielsen T. Effect of load level and muscle pain intensity on the motor control of elbow-flexion movements. Eur J Appl Physiol 2004;92(1–2):168–75.
Falla D, Gizzi L, Tschapek M, Erlenwein J, Petzke F. Reduced task-induced variations in the distribution of activity across back muscle regions in individuals with low back pain. Pain 2014;155(5):944–53.
Galati-Petrecca, M. 2008. Swiss Health Survey First findings 2007, 1sted.Berne: Federal Statistical Office; 2008.
Hoy D, Brooks P, Blyth F, Buchbinder R. The epidemiology of low back pain. Best Pract Res Clin Rheumatol 2010;24(6):769–81.
Hoozemans MJ, van der Beek AJ, Frings-Dresen MH, van der Woude LH, Van Dijk FJ. Pushing and pulling in association with low back and shoulder complaints. Occup Environ Med 2002;59(10):696–702.
Humphrey A, Nargol AF, Jones AC, Ratcliffe A, Greenough C. The value of electromyography of the lumbar paraspinal muscles in discriminating between chronic-low-back-pain sufferers and normal subjects. Eur Spine J 2005;14(2):175–84.
Lamoth CC, Meijer O, Daffertshofer A, Wuisman PJM, Beek P. Effects of chronic low back pain on trunk coordination and back muscle activity during walking: changes in motor control. Eur Spine J 2006;15(1):23–40.
Luomajoki H, Kool J, de Bruin E, Airaksinen O. Reliability of movement control tests in the lumbar spine. BMC Musculoskelet Disord 2007;8:90.
Luomajoki H, Kool J, de Bruin E, Airaksinen O. Movement control tests of the low back; evaluation of the difference between patients with low back pain and healthy controls. BMC Musculoskelet Disord 2008;9:170.
Luomajoki H, Moseley GL. Tactile acuity and lumbopelvic motor control in patients with back pain and healthy controls. Br J Sports Med 2011;45(5):437–40.
Madgwick S, Vaidyanathan R, Harrison A. An efficient orientation filter for IMU and MARG sensor arrays. Department of Mechanical Engineering, University of Bristol; 2010.
Mannion AF, Junge A, Fairbank JC, Dvorak J, Grob D. Development of a German version of the Oswestry Disability Index. Part 1: cross-cultural adaptation, reliability, and validity. Eur Spine J 2006;15(1):55–65.
Marras WS, Lavender SA, Leurgans SE, Rajulu SL, Allread WG, Fathallah FA, et al. The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders. The effects of workplace factors, trunk position, and trunk motion characteristics on risk of injury. Spine 1993;18(5):617–28.
Marras WS, Parnianpour M, Ferguson SA, Kim JY, Crowell RB, Bose S, et al. The classification of anatomic- and symptom-based low back disorders using motion measure models. Spine 1995;20(23):2531–46.
Marwan N, Kurths J. Nonlinear analysis of bivariate data with cross recurrence plots. Phys Lett A 2002;302:299–307.
Matre D, Arendt-Nielsen L, Knardahl S. Effects of localization and intensity of experimental muscle pain on ankle joint proprioception. Eur J Pain 2002;6(4):245–60.
O'Sullivan PB. Diagnosis and classification of chronic low back pain disorders: maladaptive movement and motor control impairments as underlying mechanism. Manual Ther 2005;10(4):242–55.
Price DB, Bush FM, Long S, Harkins SW. A comparison of pain measurement characteristics of mechanical visual analogue and simple numerical rating scales. Pain 1994;56(2):217–26.
Schelldorfer S, Ernst MJ, Rast FM, Bauer CM, Meichtry A, Kool J. Low back pain and postural control, effects of task difficulty on centre of pressure and spinal kinematics. Gait Posture 2015;41(1):112–8.
Silfies SP, Bhattacharya A, Biely S, Smith SS, Giszter S. Trunk control during standing reach: a dynamical system analysis of movement strategies in patients with mechanical low back pain. Gait Posture 2009;29(3):370–6.
Uri O, Pritsch M, Oran A, Liebermann DG. Upper limb kinematics after arthroscopic and open shoulder stabilization. J Shoulder Elbow Surg 2015;24(3):399–406.
Vibe Fersum K, O'Sullivan PB, Kvale A, Kouen JS. Inter-examiner reliability of a classification system for patients with non-specific low back pain. Man Ther 2009;14(5):555–61.
Webber Jr CL, Zbilut JP. Dynamical assessment of physiological systems and states using recurrence plot strategies. J Appl Physiol 1994;76(2):965–73.

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The effect of muscle fatigue and low back pain on lumbar movement variability and complexity

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ABSTRACT

Introduction: Changes in movement variability and complexity may reflect an adaptation strategy to fatigue. One unresolved question is whether this adaptation is hampered by the presence of low back pain (LBP). This study investigated if changes in movement variability and complexity after fatigue are influenced by the presence of LBP. It is hypothesised that pain free people and people suffering from LBP differ in their response to fatigue.

Methods: The effect of an isometric endurance test on lumbar movement was tested in 27 pain free participants and 59 participants suffering from LBP. Movement variability and complexity were quantified with %determinism and sample entropy of lumbar angular displacement and velocity. Generalized linear models were fitted for each outcome. Bayesian estimation of the group-fatigue effect with 95% highest posterior density intervals (95%HPDI) was performed.

Results: After fatiguing %determinism decreased and sample entropy increased in the pain free group, compared to the LBP group. The corresponding group-fatigue effects were 3.7 (95%HPDI: 2.3–7.1) and −1.4 (95%HPDI: −2.7 to −0.1). These effects manifested in angular velocity, but not in angular displacement.

Discussion: The effects indicate that pain free participants showed more complex and less predictable lumbar movement with a lower degree of structure in its variability following fatigue while participants suffering from LBP did not. This may be physiological responses to avoid overload of fatigued tissue, increase endurance, or a consequence of reduced movement control caused by fatigue.

1. Introduction

Low back pain (LBP) is one of the most common musculoskeletal disorders in industrialized countries, with a high prevalence in occupations requiring heavy repetitive physical work for the lower back (Harcombe et al., 2014; Wang et al., 2015). Heavy repetitive physical work may provoke pain by accelerating lumbar spinal disease such as cumulative trauma disorders (Solomonow, 2012) typically caused by maintaining ergonomically poor postures such as working in a stooped position with a twisted back (Holtermann et al., 2013; Roffey et al., 2010; Seidler et al., 2011; Smedley et al., 1995; Yassi and Lockhart, 2013). Further, LBP has been associated with less structured variability of lumbar movement and more structured variability of acessory lumbar movement, which were measured by percentage determinism (%DET). These might indicate early functional manifestations of LBP (Bauer et al., 2015b; Dideriksen et al., 2014). Additionally less complex lumbar postural control, quantified by sample entropy (SaEn), is correlated to increased lumbar discomfort (Søndergaard et al., 2010). This association might be explained by affected trunk neuromuscular control in people suffering from LBP (Descarreaux et al., 2007; Lamoth et al., 2006; Silfies et al., 2009; Svendsen et al., 2013).

Fatigue is a physiological short-term outcome in repetitive movements and could be a precursor to musculoskeletal disorders

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such as LBP (Rempel et al., 1992). Adequate movement variability may lead to a slower development of fatigue by distributing load across adjacent tissues, and thus maintaining task performance (Farina et al., 2008; van Dieen et al., 2003). In the presence of fatigue, the neuromuscular system rapidly searches for a new movement solution so that task performance can be preserved. This is evidenced by increased movement variabilities after fatiguing, found both at the local site of fatigue as well as the whole body (Fuller et al., 2011). Motor variability in the non-fatigued state may predict the ability to perform prolonged work (Skurvydas et al., 2010). This indicates the neuromuscular system's capability to produce sustained force during dynamic and repetitive tasks, by using variable muscle recruitment patterns. Recruitment variability, of both motor units and muscles, have a bearing on endurance during both sustained and intermittent isometric contractions of the corresponding muscles. This has been observed in trunk (van Dieen et al., 2003, 2009), and shoulder muscles (Falla and Farina, 2007; Farina et al., 2008; Palmerud et al., 1998). One unresolved question is whether these adaptations to fatigue are hampered by the presence of pain, as an impaired adaptation to fatigue might augment the development of musculoskeletal disorders. The purpose of this study was to therefore investigate if changes in movement variability and complexity after the onset of fatigue are influenced by LBP.

2. Methods

The degree of structure in the variability of lumbar movement and its complexity were assessed during a repetitive test before and immediately after an isometric endurance test to fatigue the lumbar musculature, in pain free people and people suffering from LBP. It was hypothesised that pain free people and people suffering from LBP differ in their response to fatigue. Anthropometric factors such as age, gender, or body mass index (BMI) which can influence lumbar kinematics (Consmuller et al., 2012) and the development of LBP (Heuch et al., 2015), were controlled for in the study design.

2.1. Participants

Fifty-nine participants with sub-acute or chronic LBP and 27 asymptomatic participants, aged between 18 and 65 years, were recruited from physiotherapy practices, the university campus and through newspaper advertisements. To be eligible participants with LBP had to fulfil the following inclusion criteria:

- a current episode of sub-acute or chronic non-specific LBP that persisted for four weeks or longer;
- a mean LBP intensity of $\geq$1 on the numeric rating scale (NRS) over the last four weeks;
- a moderate disability defined as an Oswestry-disability-index (ODI) > 8%;
- a low level of psychosocial risk factors defined with less than four points on the psychosocial subscale of the STarT Back screening tool (Mannion et al., 2006).

Exclusion criteria were:

- specific LBP;
- vertigo or disturbance of the equilibrium;
- systemic diseases (diabetes, tumours);
- pain in other areas of the body (neck, head, thoracic spine, or arms);
- complaints, injury, or surgery of the legs (hips to feet) within the last six months;
- medication affecting postural control (e.g. anti-depressants);
- pregnancy.

In addition the asymptomatic participants were excluded if they had a LBP episode during the preceding three months. The study was conducted according to the declaration of Helsinki, and approved by the local ethics committee (KEK-ZH-2011-0522). Participants provided their written informed consent.

2.2. Experimental procedures

Participants attended one measurement session and performed one "Repeated Flexion and Extension" test (Fig. 2) (Bauer et al., 2015a), pre and post fatiguing of the lumbar musculature. The test consisted of twenty cycles, of three seconds duration, starting in upright sitting. The duration of the complete test was 60 s. During each cycle the participants were asked to flex and extend their lumbar spine and hip as far as possible whilst adhering to the timed window. They were guided with a metronome set at 80 bpm and were instructed to flex and extend their lumbar spine within two beats respectively. The participants were fixed at their thighs with two belts to prevent unintended movement of their pelvis and thighs. The test was performed one time pre and post fatiguing, respectively. Prior to the pre-test the participants received standardized oral instructions by one of the examiners and visual instructions in a video. In case of poor initial performance these instructions were repeated up to three times and the test was demonstrated by one examiner. The participants performed the post-test immediately after completing an isometric endurance test (Biering-Sorensen, 1984) to fatigue the lumbar musculature.

The isometric endurance test is described elsewhere (Biering-Sorensen, 1984) and only briefly summarized here. The participants were lying in prone position on a physiotherapy bench that was tilted 45° downwards from the pelvis. The participants lower legs and thighs were fixated with two belts. They were instructed to lift their upper body from the bench and maintain it unsupported for as long as possible. When the participants could not maintain this any longer, or one investigator noticed that their upper body touched the bench, the test was ended and the post-test was performed immediately after. All participants received standardized verbal encouragement during the endurance test.

2.3. Movement analysis

2.3.1. Sensor placement and data processing

Lumbar spine movement was measured by an inertial measurement unit (IMU) system, with IMUs placed over the level of the second sacral and first lumbar vertebra (Ernst et al., 2013) (Fig. 1). The IMU system has been shown to provide concurrently

Fig. 1. Experimental setup: IMUs were placed on the level of sacrum (S2) and L1 (L1).

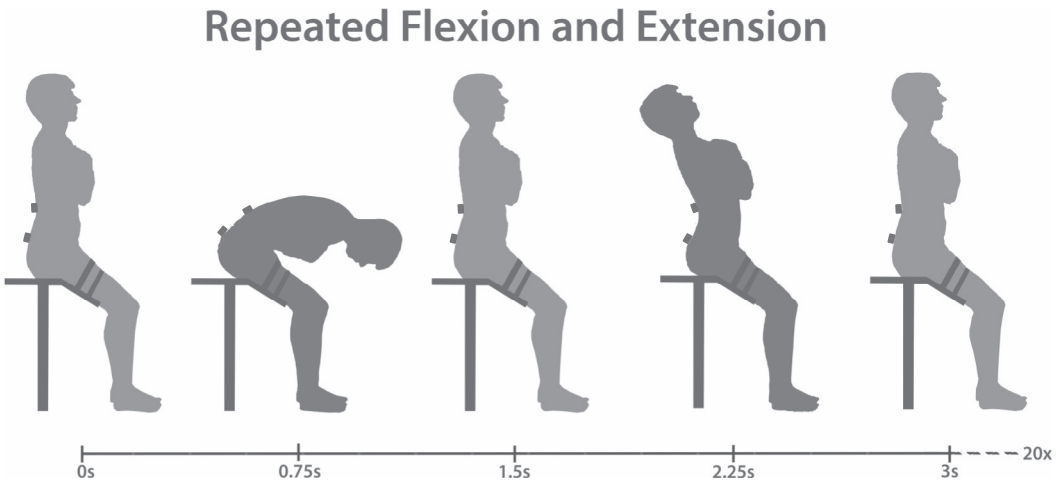

Fig. 2. Repeated flexion and extension test.

valid estimates of spinal kinematics (Bauer et al., 2015a). The sensors of the IMU system (Valedo®Motion, Hocoma AG, Volketswil, Switzerland) include a tri-axial gyroscope, magnetometer, and accelerometer. Movement data were recorded with a sampling frequency of 200 Hz using customized software (Valedo®Research, Hocoma AG). The raw data from the IMUs were transformed into quaternions (Madgwick et al., 2010) and filtered using a second-order zero-phase low-pass Butterworth filter (6 Hz cut-off frequency). Segmental kinematics were calculated using the tilt/twist formulation to prevent rotational singularities (Crawford et al., 1999) with sagittal plane defined by the global coordinate system. All outcome variables were derived from the flexion/extension angle, where flexion is positive and extension is negative. An angle of zero degrees is defined as alignment of the IMUs. Angular velocity was calculated using the first derivative of the angular displacement data. A complete description of the data processing from raw data to tilt/twist angles is described elsewhere (Bauer et al., 2015a).

2.3.2. Outcomes

Percentage determinism (%DET) and sample entropy (SaEn) were calculated from angular displacement and velocity data (Fig. 3). %DET and SaEn describe different aspects of a time series signal: %DET indicates the predictability of a signal by providing an indication of the structure in variability. Lower %DET implies a lower degree of structure in the signals variability (Webber and Zbilut, 1994). SaEn is a measure of complexity. Lower SaEn indicates that a signal is less complex (Richman and Moorman, 2000). As such the future state of the time-series is less complex and more predictable. To calculate both parameters %DET and SaEn, movement data were projected into a phase space by taking time-delayed samples from the movement data. The time-delayed samples represent movement patterns which can be visualized as points in the phase-space plot. The vectors consisting of the time-delayed samples are called embedding vectors.

Recurrence quantification was used to calculate %DET. In recurrence quantification analysis, similar movement patterns are located close to each other, and form a cluster of recurrent points ($R_{i,j}$) (Webber and Zbilut, 1994). The similarity of movement patterns is quantified by calculating the Euclidean distances between the embedding vectors. In this study, the phase-space reconstruction was undertaken separately for angular displacement and velocity data by using the set of parameters specified in Table 1. All $R_{i,j}$:s were subsequently transferred into a $N \times N$-sized recurrence plot (RP) (Fig. 4) with N being the number of points in the RP. *%DET* is the amount of $R_{i,j}$ that form diagonal lines (i.e. are sequential to each other in time) of a prespecified minimal length (*lmin*) given in Table 1. The %DET is expressed as:

$$\%DET = \frac{\sum_{l=lmin}^{lmax} l * P(l)}{\sum_{l=1}^{lmax} l * P(l)} * 10^2$$

with l being the length of the diagonal lines, *lmax* the maximal possible length of the diagonal lines, and $P(l)$ being the number of diagonal lines of length l.

To calculate SaEn, the similarity of movement patterns was quantified by calculating the distances between the embedding vectors as the maximum difference of their corresponding scalar components. The length of created embedding vectors was m and m + 1. SaEn was calculated as:

$$SaEn = -\ln\left(\frac{\Phi^{m+1}(r)}{\Phi^{m}(r)}\right)$$

with Φ being the probability that two embedding vectors are similar in comparison to previously defined tolerance r (Richman and Moorman, 2000). All data processing and calculations were done using Matlab 2012b® (Mathworks, USA), with %DET code from University of Potsdam, Germany (Marwan and Kurths, 2002) and SaEn code from Nanyang Technological University, Singapore (Lee, 2012).

2.4. Statistical analysis

For each outcome a linear mixed model for observation Y_{ijk} (kth participant in the ith group at time j) was modelled with

$$Y_{ijk} = \mu + \alpha_i + \beta_j + (\alpha\beta)_{ij} + U_{k(i)} + \beta_{covariates} + \varepsilon_{ijk}$$

with α_i as the i_{th} group (LBP/painfree) effect, β_j as the j_{th} fatigue effect, $(\alpha\beta)_{ij}$as the ij_{th} group-fatigue effect, $U_{k(i)}$ as the effect of subject k nested in group i, $\beta_{covariates}$ as the effect of age, gender and BMI, and ε_{ijk} as measurement error. We assume that $U_{ik} \sim N(0, v^2)$ with v^2 as the between-subject variance and ε_{ijk} $N(0, \tau^2)$ withτ^2 as the within-subject variance. Bayesian estimation of the model parameters was performed by using uninformative priors on the model parameters. We used Monte-Carlo-Markov-Chain (MCMC) algorithms with Gibbs sampling (Plummer, 2003) to sample from the

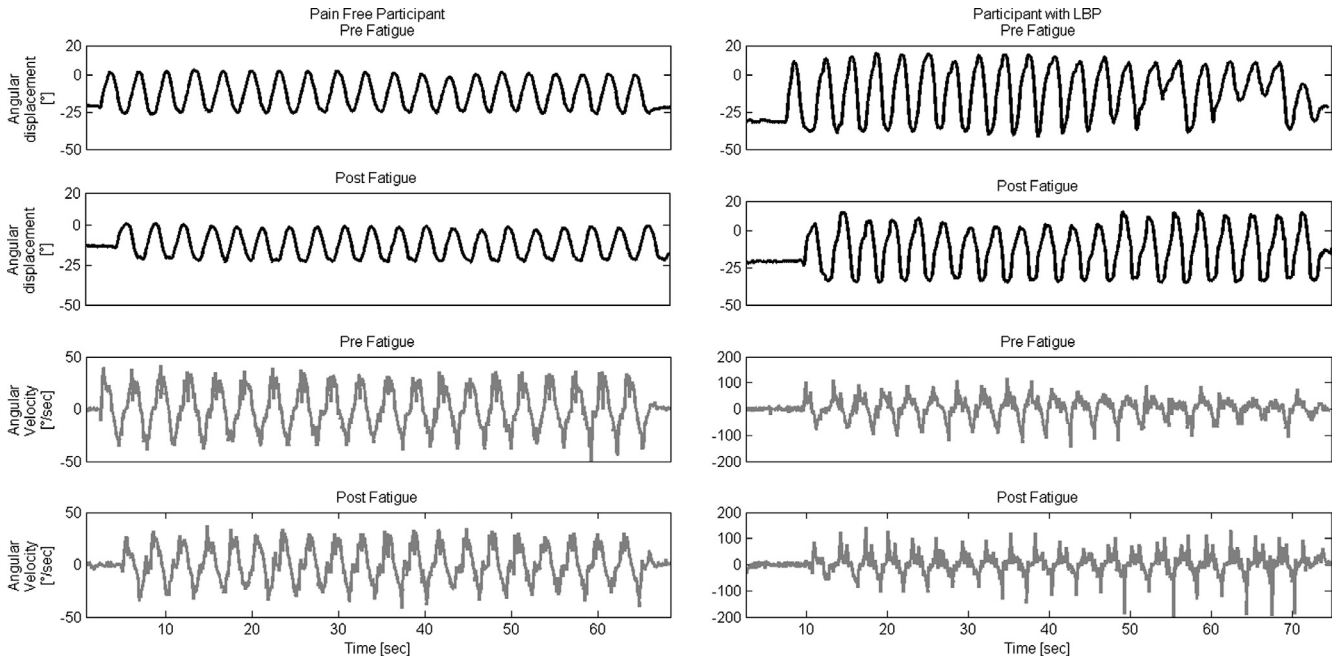

Fig. 3. Angular displacement and velocity data pre and post fatigue. The left column shows angular displacement and velocity of a pain free participant, the right column a participant with low back pain. Determinism decreases, while sample entropy increases in the pain free participant.

Table 1
Input parameters used in recurrence quantification analysis and sample entropy calculation.

	Recurrence quantification analysis					Sample entropy	
	Delay	Embedding dimension	Distance	lmin	Tolerance	Embedding dimension	Tolerance
Angular displacement	35	2	Euclidian	150	1.3*SD	2	0.2*SD
Angular velocity	21	2	Euclidian	150	1.3*SD	2	0.2*SD

lmin - minimal length of diagonal line; SD – standard deviation.
The delays were estimated using mutual information analysis. The first minimum of mutual information was defined as the optimal delay. The embedding dimensions were estimated by calculating the correlation dimension with different embedding dimensions. The optimal value of the embedding dimension was chosen as the starting point where the correlation dimension did not increase significantly although increasing the embedding dimension. The tolerance for determining a recurrent point in the reconstructed state-space was calculated from the standard deviation of the phase space trajectory. The optimal minimal length of diagonal line was chosen after visual inspection of the recurrence plots.

posterior distributions. We created point estimates of group-fatigue effects and 95% Highest Posterior Density intervals (95%HPDI) for the parameters in an MCMC sample. The repeatability of %DET and SaEn is presented in supplementary file 1. A group-fatigue effect would indicate that one group responds differently to fatigue compared to the other. The probability that the magnitude of this effect is within the borders of the 95%HPDI is 0.95. The HPDI is thus comparable to the confidence interval in frequentist statistics. Statistical analysis was conducted using R (R Foundation for statistical computing, Austria).

3. Results

The descriptive characteristics, group-fatigue effects and corresponding 95%HPDI are presented in Table 2. Pain free participants showed more complex and less predictable lumbar movement, with a lower degree of structure in its variability while participants suffering from LBP did not. This effect was only observed, after an isometric endurance test, in lumbar angular velocity, but not in angular displacement. %DET and SaEn of angular displacement did not show a group-fatigue effect, indicated by the 95%HPDI crossing 0. Angular velocity %DET decreased and SaEn increased more distinctively in the painfree group, indicating a group-fatigue effect, with the 95%HPDI not crossing 0. The group-fatigue effects on %DET and SaEn are illustrated in Fig. 5. Additionally, angular velocity %DET decreased −0.2, while SaEn increased 0.001 per year of age. Gender and BMI had no effect on %DET and SaEn.

4. Discussion

This study revealed that pain free participants displayed more complex and less predictable lumbar movement with a lower degree of structure in its variability following an isometric endurance test to fatigue lumbar musculature while participants suffering from LBP did not. This effect was only observed in lumbar angular velocity during a repetitive movement test, but not in lumbar angular displacement.

These findings indicate that the presence of LBP influences a person's response to fatigue. Pain free persons might adapt to fatigue by showing more complex and less predictable lumbar movement with a lower degree of structure in its variability which is theorized as a strategy to reduce load on fatiguing tissues, while preserving task performance. Evidence suggesting that changes in movement variability may help in preserving performance during a fatiguing task has been reported in repetitive throwing (Forestier and Nougier, 1998; Huffenus et al., 2006), repetitive reaching (Fuller et al., 2011), cross-country skiing (Cignetti et al.,

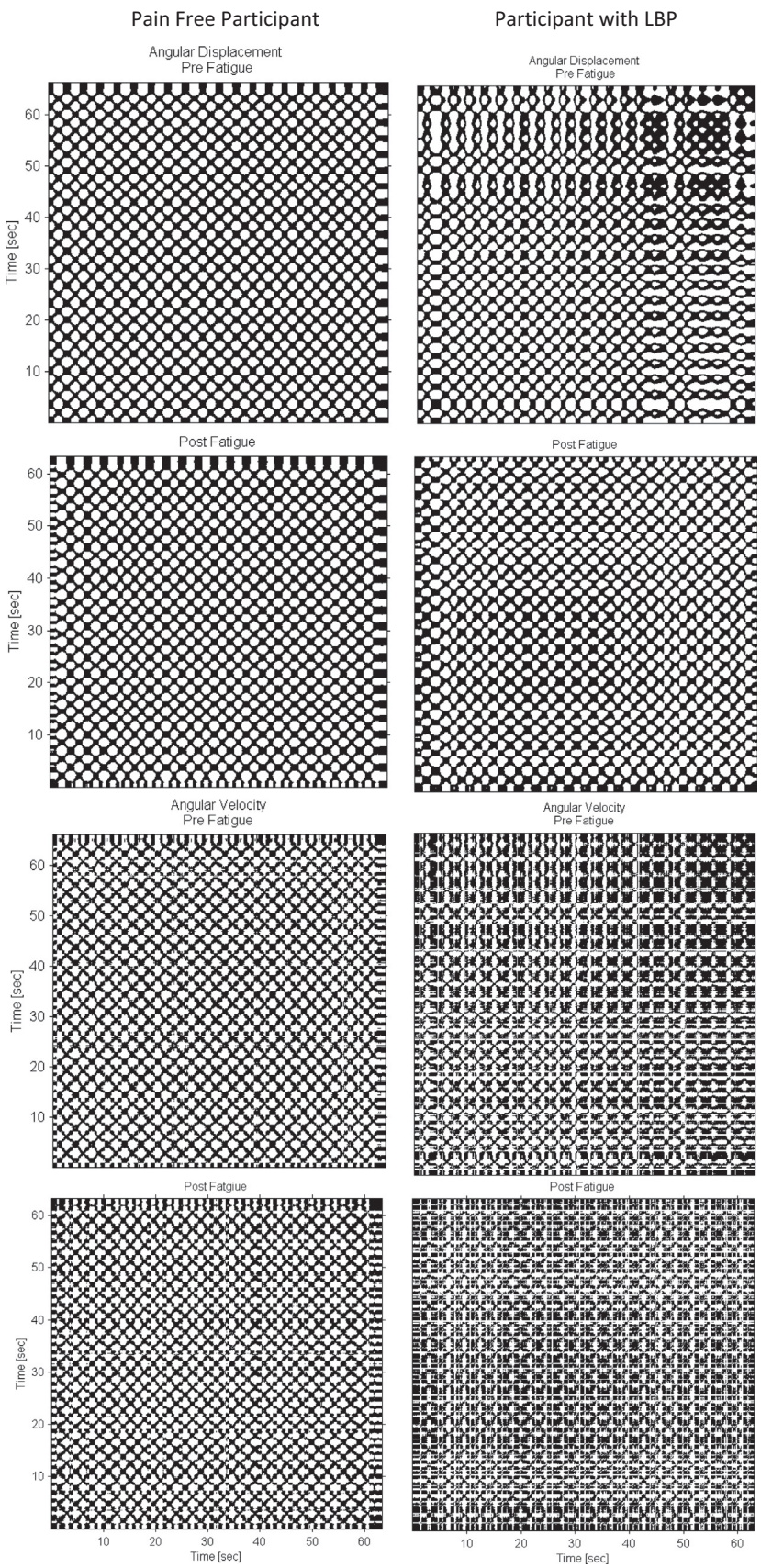

Fig. 4. Recurrence plots of angular displacement and velocity data pre and post fatigue. The left column shows angular displacement and velocity of a painfree participant in, the right column a participant with low back pain. Determinism decreases, post fatigue, in the pain free participant.

Table 2
Descriptive statistics and group-fatigue effect.

Group	n	Gender m/f	Age years	BMI (kg/m^2)	Time to fatigue (seconds)	Fatigue	RoM (°)	Angular displacement		Angular velocity	
								%DET	SaEn	%DET	SaEn
Painfree	27	12/15	39.6 (±11.6)	22.7 (±2.8)	161.8 (±63.3)	pre	60.0 (±14.9)	62.8 (±2.9)	2.4 (±0.3)	36.2 (±8.8)	9.7 (±2.4)
						post	61.7(±21.8)	63.3 (±2.9)	2.5 (±0.5)	32.1 (±9.3)	11.2 (±3.6)
LBP	59	30/29	39.1 (±12.8)	24.0 (±3.6)	153.8 (±56.8)	pre	59.5 (±19.1)	64.6 (±3.9)	2.4 (±0.4)	34.9 (±8.1)	11.7 (±3.4)
						post	61.6 (±22.4)	64.4 (±4.2)	2.5 (±0.6)	34.4 (±9.8)	11.8 (±3.7)
Statistics											
Group-fatigue effect								−0.6	−0.1	**3.7**	**−1.4**
95%HPDI								−2.4 to 1.0	−0.4 to 0.2	**2.3 to 7.1**	**−2.7 to −0.1**

95% HPDI - 95% highest posterior density interval; BMI – body mass index; %DET – % of determinism; LBP –low back pain; NRS- numeric pain rating scale; RoM – range of motion lumbar spine; SaEn - sample entropy.
Results are provided as mean (±standard deviation); Bold numbers indicate the 95% HDPI not crossing 0. SaEn values are expressed as $*10^2$.

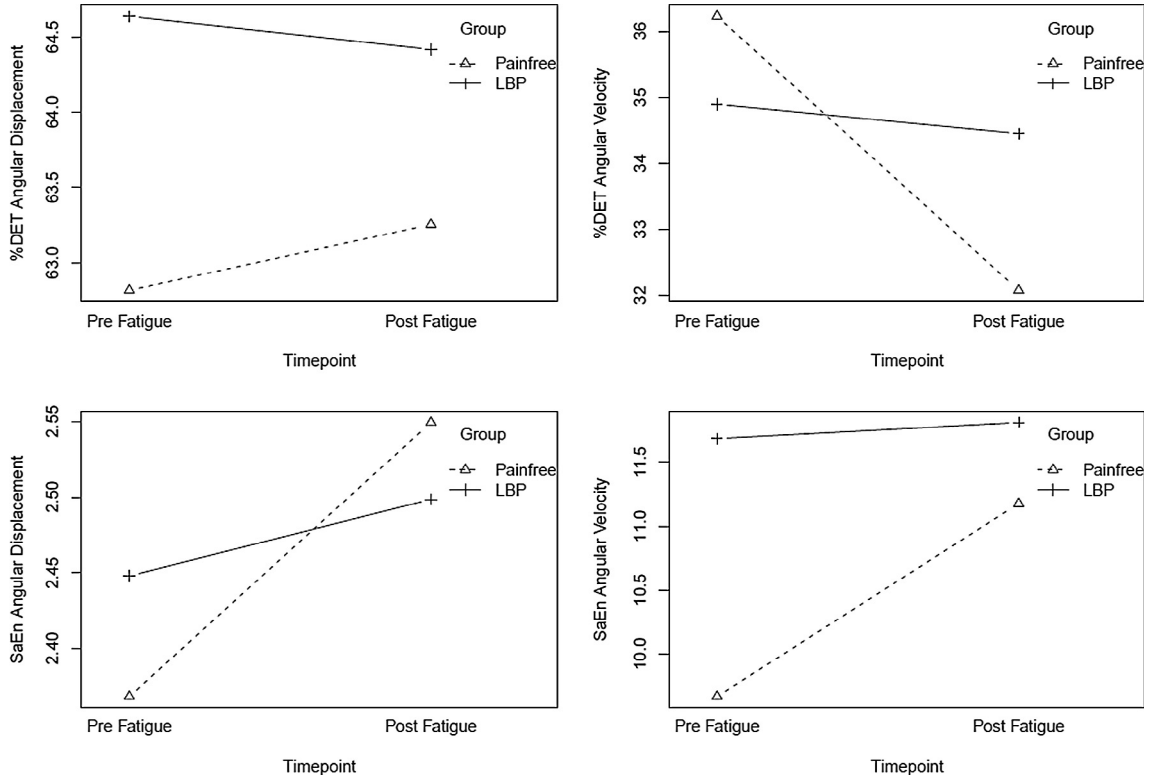

Fig. 5. Interaction plots for the primary outcomes pre and post fatigue. Abbreviations: % DET – percentage determinism; LBP – low back pain; SaEn – Sample Entropy

2009), hammering tasks (Cote et al., 2008, 2005), and repeated elbow flexion/extension for tracking a target (Selen et al., 2007). On the other hand, people suffering from LBP may be unable to adapt their movement strategy by making use of the musculoskeletal systems redundancy, thus accumulating load on fatiguing tissues. This could be explained by pain induced changes in muscle and motor unit recruitment patterns that have been observed under experimental pain conditions (Falla et al., 2009; Farina et al., 2012; Muceli et al., 2014; Yavuz et al., 2015).

On the contrary the present findings might indicate that pain free participants were able to control their lumbar movement before the onset of fatigue, but lost control of lumbar movement after the onset of fatigue, thereupon resembling participants suffering from LBP. This is indicated by the less complex and more predictable lumbar movement with a higher degree of structure in its variability that pain free participants showed during pretest. This interpretation would imply that more structured movement variability might actually be beneficial and represent better movement control during repetitive tasks. This hypothesis is supported by findings that less structured lumbar movement variability is associated with increased LBP intensity (Bauer et al., 2015b). Furthermore it might indicate an inability of the lumbar para-spinal muscles to stabilize the lumbar spine in the neutral zone. This hypothesis should be addressed in a future study on lumbar muscle recruitment patterns.

The optimal, or healthy, complexity and degree of structure in variability needed to prevent lumbar musculoskeletal disorders is unknown. The presence of variability in movement patterns may represent the underlying physiological capability to make flexible adaptations to everyday stressors placed on the neuromuscular system (Lipsitz, 2002; Lipsitz and Goldberger, 1992). In contrast, the presence of ergonomically poor movement patterns might increase the measured movement variability, thus elevating tissue loading. Therefore, a non-linear or U-shaped relationship between structure, complexity and disease is hypothesised (Stergiou and Decker, 2011), and reported by previous studies: More structured

variability of lumbar movement (Bauer et al., 2015b) but less structured variability of acessory lumbar movement (Dideriksen et al., 2014) were associated with LBP, which might indicate early functional manifestations of LBP. Larger and smaller sizes of arm and trunk movement variability were found during simulated low-load repetitive work in people suffering from acute and sub-acute or chronic pain respectively (Madeleine, 2010).

A limitation of the present study is that the true size of the effect of fatigue on the complexity of lumbar movement and the degree of structure in its variability remains unknown since this study did not include a measure of lumbar muscular fatigue. This indicates the need to record EMG data to supplement kinematic analyses and analyse changes in %DET of EMG data pre and post fatigue. Increases in %DET have been reported in the presence of fatigue; which reflects higher periodicity (Felici et al., 2001). The inclusion of a measure of lumbar strength could have helped to quantify if both groups were equally influenced by the protocol. Future studies might consider additional lateral flexion, and rotation angles. They were not analysed due to the IMU systems limited concurrent validity when measuring lateral flexion or rotation movements of small magnitude during large flexion extension movements (Bauer et al., 2015a). Furthermore, the current study design does not enable to say if the difference in response to fatigue is a cause or a result of LBP. Still, this study demonstrated that changes in lumbar movement velocity after an isometric endurance test are influenced by the presence of LBP.

This study shows that painfree people differ in their kinematic response to fatigue compared to people suffering from LBP. Future studies should combine measures of lumbar kinematics with measures of cortical representation and muscle recruitment patterns to improve our understanding of the underlying mechanisms. Further, those measures should be combined with work performance measures, to evaluate whether the observed changes following fatigue preserve an individual's task performance. Eventually, longitudinal prospective studies should investigate the development of LBP and corresponding changes in complexity of lumbar movement and structure of its variability, to assess whether these changes are a cause or consequence of LBP.

5. Conclusions

The effect of fatigue on the complexity of lumbar movement and the degree of structure in its variability was analysed in pain free participants and participants suffering from LBP, and controlled for the effect of age, gender and BMI. A group-fatigue effect on lumbar movement velocity was found, meaning that pain free people show more complex and less predictable lumbar movement velocity with a lower degree of structure in its variability, while people suffering from LBP do not. These findings indicate that the presence of LBP influences a participant's response to an isometric endurance test.

Conflict of interest

The authors declare that they have no conflict of interest.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jelekin.2017.02.003.

References

Bauer, C.M., Rast, F.M., Ernst, M.J., Kool, J., Oetiker, S., Rissanen, S.M., et al., 2015a. Concurrent validity and reliability of a novel wireless inertial measurement system to assess trunk movement. J. Electromyogr. Kinesiol. 25, 782–790.
Bauer, C.M., Rast, F.M., Ernst, M.J., Oetiker, S., Meichtry, A., Kool, J., et al., 2015b. Pain intensity attenuates movement control of the lumbar spine in low back pain. J. Electromyogr. Kinesiol. 25, 919–927.
Biering-Sorensen, F., 1984. Physical measurements as risk indicators for low-back trouble over a one-year period. Spine 9, 106–119.
Cignetti, F., Schena, F., Rouard, A., 2009. Effects of fatigue on inter-cycle variability in cross-country skiing. J. Biomech. 42, 1452–1459.
Consmuller, T., Rohlmann, A., Weinland, D., Druschel, C., Duda, G.N., Taylor, W.R., 2012. Velocity of lordosis angle during spinal flexion and extension. PLoS One 7, e50135.
Cote, J.N., Feldman, A.G., Mathieu, P.A., Levin, M.F., 2008. Effects of fatigue on intermuscular coordination during repetitive hammering. Motor Control 12, 79–92.
Cote, J.N., Raymond, D., Mathieu, P.A., Feldman, A.G., Levin, M.F., 2005. Differences in multi-joint kinematic patterns of repetitive hammering in healthy, fatigued and shoulder-injured individuals. Clin. Biomech. 20, 581–590.
Crawford, N.R., Yamaguchi, G.T., Dickman, C.A., 1999. A new technique for determining 3-D joint angles: the tilt/twist method. Clin. Biomech. 14, 153–165.
Descarreaux, M., Lalonde, C., Normand, M.C., 2007. Isometric force parameters and trunk muscle recruitment strategies in a population with low back pain. J. Manipulative Physiol. Ther. 30, 91–97.
Dideriksen, J.L., Gizzi, L., Petzke, F., Falla, D., 2014. Deterministic accessory spinal movement in functional tasks characterizes individuals with low back pain. Clin. Neurophysiol. 125, 1663–1668.
Ernst, M.J., Rast, F.M., Bauer, C.M., Marcar, V.L., Kool, J., 2013. Determination of thoracic and lumbar spinal processes by their percentage position between C7 and the PSIS level. BMC Res. Notes 6, 58.
Falla, D., Arendt-Nielsen, L., Farina, D., 2009. The pain-induced change in relative activation of upper trapezius muscle regions is independent of the site of noxious stimulation. Clin. Neurophysiol. 120, 150–157.
Falla, D., Farina, D., 2007. Periodic increases in force during sustained contraction reduce fatigue and facilitate spatial redistribution of trapezius muscle activity. Exp. Brain Res. 182, 99–107.
Farina, D., Leclerc, F., Arendt-Nielsen, L., Buttelli, O., Madeleine, P., 2008. The change in spatial distribution of upper trapezius muscle activity is correlated to contraction duration. J. Electromyogr. Kinesiol. 18, 16–25.
Farina, D., Negro, F., Gizzi, L., Falla, D., 2012. Low-frequency oscillations of the neural drive to the muscle are increased with experimental muscle pain. J. Neurophysiol. 107, 958–965.
Felici, F., Rosponi, A., Sbriccoli, P., Filligoi, G.C., Fattorini, L., Marchetti, M., 2001. Linear and non-linear analysis of surface electromyograms in weightlifters. Eur. J. Appl. Physiol. 84, 337–342.
Forestier, N., Nougier, V., 1998. The effects of muscular fatigue on the coordination of a multijoint movement in human. Neurosci. Lett. 252, 187–190.
Fuller, J.R., Fung, J., Cote, J.N., 2011. Time-dependent adaptations to posture and movement characteristics during the development of repetitive reaching induced fatigue. Exp. Brain Res. 211, 133–143.
Harcombe, H., Herbison, G.P., McBride, D., Derrett, S., 2014. Musculoskeletal disorders among nurses compared with two other occupational groups. Occup. Med. 64, 601–607.
Heuch, I., Heuch, I., Hagen, K., Zwart, J.A., 2015. A comparison of anthropometric measures for assessing the association between body size and risk of chronic low back pain: the HUNT study. PLoS One 10, e0141268.
Holtermann, A., Clausen, T., Jorgensen, M.B., Burdorf, A., Andersen, L.L., 2013. Patient handling and risk for developing persistent low-back pain among female healthcare workers. Scand. J. Work Environ. Health 39, 164–169.
Huffenus, A.F., Amarantini, D., Forestier, N., 2006. Effects of distal and proximal arm muscles fatigue on multi-joint movement organization. Exp. Brain Res. 170, 438–447.
Lamoth, C.C., Meijer, O., Daffertshofer, A., Wuisman, P.J.M., Beek, P., 2006. Effects of chronic low back pain on trunk coordination and back muscle activity during walking: changes in motor control. Eur. Spine J. 15, 23–40.
Lee, K., 2012. Sample Entropy. MathWorks, File Exchange; <https://www.mathworks.com/matlabcentral/fileexchange/35784-sample-entropy>.
Lipsitz, L.A., 2002. Dynamics of stability: the physiologic basis of functional health and frailty. J. Gerontol. A Biol. Sci. Med. Sci. 57, B115–B125.
Lipsitz, L.A., Goldberger, A.L., 1992. Loss of 'complexity' and aging. Potential applications of fractals and chaos theory to senescence. Jama 267, 1806–1809.
Madeleine, P., 2010. On functional motor adaptations: from the quantification of motor strategies to the prevention of musculoskeletal disorders in the neck-shoulder region. Acta Physiol. 199 (Suppl. 679), 1–46.

Madgwick, S., Vaidyanathan, R., Harrison, A., 2010. An Efficient Orientation Filter for IMU and MARG Sensor Arrays. Department of Mechanical Engineering, University of Bristol.
Mannion, A.F., Junge, A., Fairbank, J.C., Dvorak, J., Grob, D., 2006. Development of a German version of the oswestry disability index. Part 1: cross-cultural adaptation, reliability, and validity. Eur. Spine J. 15, 55–65.
Marwan, N., Kurths, J., 2002. Nonlinear analysis of bivariate data with cross recurrence plots. Phys. Lett. A 302, 299–307.
Muceli, S., Falla, D., Farina, D., 2014. Reorganization of muscle synergies during multidirectional reaching in the horizontal plane with experimental muscle pain. J. Neurophysiol. 111, 1615–1630.
Palmerud, G., Sporrong, H., Herberts, P., Kadefors, R., 1998. Consequences of trapezius relaxation on the distribution of shoulder muscle forces: an electromyographic study. J. Electromyogr. Kinesiol. 8, 185–193.
Plummer, M., 2003. JAGS: a program for analysis of Bayesian graphical models using Gibbs sampling. In: Kurt Hornik, F.L.A.Z. (Ed.), Proceedings of the 3rd International Workshop on Distributed Statistical Computing (DSC 2003).
Rempel, D.M., Harrison, R.J., Barnhart, S., 1992. Work-related cumulative trauma disorders of the upper extremity. Jama 267, 838–842.
Richman, J.S., Moorman, J.R., 2000. Physiological time-series analysis using approximate entropy and sample entropy. Am. J. Physiol. Heart Circ. Physiol. 278, H2039–H2049.
Roffey, D.M., Wai, E.K., Bishop, P., Kwon, B.K., Dagenais, S., 2010. Causal assessment of workplace manual handling or assisting patients and low back pain: results of a systematic review. Spine J. 10, 639–651.
Seidler, A., Euler, U., Bolm-Audorff, U., Ellegast, R., Grifka, J., Haerting, J., et al., 2011. Physical workload and accelerated occurrence of lumbar spine diseases: risk and rate advancement periods in a German multicenter case-control study. Scand. J. Work Environ. Health 37, 30–36.
Selen, L.P., Beek, P.J., van Dieen, J.H., 2007. Fatigue-induced changes of impedance and performance in target tracking. Exp. Brain Res. 181, 99–108.
Silfies, S.P., Bhattacharya, A., Biely, S., Smith, S.S., Giszter, S., 2009. Trunk control during standing reach: a dynamical system analysis of movement strategies in patients with mechanical low back pain. Gait Posture 29, 370–376.
Skurvydas, A., Brazaitis, M., Kamandulis, S., Sipaviciene, S., 2010. Peripheral and central fatigue after muscle-damaging exercise is muscle length dependent and inversely related. J. Electromyogr. Kinesiol. 20, 655–660.
Smedley, J., Egger, P., Cooper, C., Coggon, D., 1995. Manual handling activities and risk of low back pain in nurses. Occup. Environ. Med. 52, 160–163.
Solomonow, M., 2012. Neuromuscular manifestations of viscoelastic tissue degradation following high and low risk repetitive lumbar flexion. J. Electromyogr. Kinesiol. 22, 155–175.
Stergiou, N., Decker, L.M., 2011. Human movement variability, nonlinear dynamics, and pathology: is there a connection? Hum. Mov. Sci. 30, 869–888.
Svendsen, J.H., Svarrer, H., Laessoe, U., Vollenbroek-Hutten, M., Madeleine, P., 2013. Standardized activities of daily living in presence of sub-acute low-back pain: a pilot study. J. Electromyogr. Kinesiol. 23, 159–165.
van Dieen, J.H., Selen, L.P., Cholewicki, J., 2003. Trunk muscle activation in low-back pain patients, an analysis of the literature. J. Electromyogr. Kinesiol. 13, 333–351.
van Dieen, J.H., Westebring-van der Putten, E.P., Kingma, I., de Looze, M.P., 2009. Low-level activity of the trunk extensor muscles causes electromyographic manifestations of fatigue in absence of decreased oxygenation. J. Electromyogr. Kinesiol. 19, 398–406.
Wang, S.Y., Liu, L.C., Lu, M.C., Koo, M., 2015. Comparisons of musculoskeletal disorders among ten different medical professions in Taiwan: a nationwide, population-based study. PLoS One 10, e0123750.
Webber Jr., C.L., Zbilut, J.P., 1994. Dynamical assessment of physiological systems and states using recurrence plot strategies. J. Appl. Physiol. 76, 965–973.
Yassi, A., Lockhart, K., 2013. Work-relatedness of low back pain in nursing personnel: a systematic review. Int. J. Occup. Environ. Health 19, 223–244.
Yavuz, U.S., Negro, F., Falla, D., Farina, D., 2015. Experimental muscle pain increases variability of neural drive to muscle and decreases motor unit coherence in tremor frequency band. J. Neurophysiol. 114, 1041–1047.
Søndergaard, K.H.E., Olesen, C.G., Søndergaard, E.K., de Zee, M., Madeleine, P., 2010. The variability and complexity of sitting postural control are associated with discomfort. J. Biomech. 43, 1997–2001.

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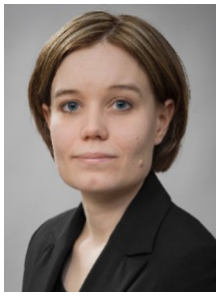

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