University of Potsdam

Faculty of Human Sciences

Department of Sport and Health Science

Professorship of Sports Medicine and Sports Orthopaedics

Trunk loading and Back Pain

- Three-dimensional motion analysis and evaluation of neuromuscular reflex activity in response to sudden and continuous loading -

An academic thesis submitted to

the Faculty of Human Sciences of the University of Potsdam

for the degree

Doctor of Philosophy (Dr. phil.)

by

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Date of submission: 26.08.2016
Date of defence: 11.01.2017

Published online at the
Institutional Repository of the University of Potsdam:
URN urn:nbn:de:kobv:517-opus4-102428
http://nbn-resolving.de/urn:nbn:de:kobv:517-opus4-102428

Für

Opa

Dedicated to

Grandpa

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Abstract - English

An essential function of the trunk is the compensation of external forces and loads in order to guarantee stability. Stabilising the trunk during sudden, repetitive loading in everyday tasks, as well as during performance is important in order to protect against injury. Hence, reduced trunk stability is accepted as a risk factor for the development of back pain (BP). An altered activity pattern including extended response and activation times as well as increased co-contraction of the trunk muscles as well as a reduced range of motion and increased movement variability of the trunk are evident in back pain patients (BPP). These differences to healthy controls (H) have been evaluated primarily in quasi-static test situations involving isolated loading directly to the trunk. Nevertheless, transferability to everyday, dynamic situations is under debate. Therefore, the aim of this project is to analyse 3-dimensional motion and neuromuscular reflex activity of the trunk as response to dynamic trunk loading in healthy (H) and back pain patients (BPP).

A measurement tool was developed to assess trunk stability, consisting of dynamic test situations. During these tests, loading of the trunk is generated by the upper and lower limbs with and without additional perturbation. Therefore, lifting of objects and stumbling while walking are adequate represents. With the help of a 12-lead EMG, neuromuscular activity of the muscles encompassing the trunk was assessed. In addition, three-dimensional trunk motion was analysed using a newly developed multi-segmental trunk model. The set-up was checked for reproducibility as well as validity. Afterwards, the defined measurement set-up was applied to assess trunk stability in comparisons of healthy and back pain patients.

Clinically acceptable to excellent reliability could be shown for the methods (EMG/kinematics) used in the test situations. No changes in trunk motion pattern could be observed in healthy adults during continuous loading (lifting of objects) of different weights. In contrast, sudden loading of the trunk through perturbations to the lower limbs during walking led to an increased neuromuscular activity and ROM of the trunk. Moreover, BPP showed a delayed muscle response time and extended duration until maximum neuromuscular activity in response to sudden walking perturbations compared to healthy controls. In addition, a reduced lateral flexion of the trunk during perturbation could be shown in BPP.

It is concluded that perturbed gait seems suitable to provoke higher demands on trunk stability in adults. The altered neuromuscular and kinematic compensation pattern in back pain patients (BPP) can be interpreted as increased spine loading and reduced trunk stability in patients. Therefore, this novel assessment of trunk stability is suitable to identify deficits in BPP. Assignment of affected BPP to therapy interventions with focus on stabilisation of the trunk aiming to improve neuromuscular control in dynamic situations is implied. Hence, sensorimotor training (SMT) to enhance trunk stability and compensation of unexpected sudden loading should be preferred.

Abstract – German

Eine ausgeprägte Rumpfstabilität gilt als vorteilhaft, um den Rumpf bei repetitiven und plötzlich auftretenden hohen Lasten sowohl in Alltagssituationen, am Arbeitsplatz als auch während Training- oder Wettkampfbelastungen im Sport zu stabilisieren und vor Beschwerden bzw. Verletzungen zu schützen. Eine reduzierte Rumpfstabilität wird daher als Risikofaktor für die Entwicklung von Rückenschmerzen angenommen. Eine veränderte Aktivität (verlängerte Reaktionszeit, verlängerte Aktivierungszeiten, erhöhte Ko-Kontraktion) der rumpfumgreifenden Muskulatur sowie ein reduziertes Bewegungsausmaß und eine erhöhte -variabilität des Rumpfes bei Rückenschmerzpatienten sind evident. Diese Unterschiede sind hauptsächlich in quasi-statischen Belastungssituationen mit isolierter Lasteinwirkung direkt am Rumpf nachgewiesen. Eine Übertragbarkeit auf alltagsnahe und dynamische Belastungssituationen ist jedoch kritisch zu hinterfragen. Ziel des Dissertationsprojektes ist die Entwicklung und Validierung eines Diagnostiktools zur Erhebung der Rumpfstabilität in dynamischen Belastungssituationen bei Gesunden und Rückenschmerzpatienten.

Für die Erfassung der Rumpfstabilität wurde ein Mess-Verfahren bestehend aus dynamischen Belastungssituationen, in denen die Lasten über die Extremitäten generiert werden (Heben von Lasten, Perturbation im Gang), mit und ohne Perturbation entwickelt. Mit Hilfe eines 12-Kanal-EMGs wurde die neuromuskuläre Aktivität der rumpfumgreifenden Muskulatur erfasst. Zusätzlich wurde die 3-dimensonale Rumpfbewegung über ein neu entwickeltes multi-segmentales Rückenmodell analysiert. Dieses Methodensetup wurde auf Reproduzierbarkeit sowie Validität überprüft. Im Anschluss erfolgte eine Anwendung des definierten Diagnostiktools zur Erfassung der Rumpfstabilität im Vergleich von Probanden mit und ohne Rückenschmerzen.

Eine klinisch akzeptable bis hervorragende Reliabilität konnte für die verwendeten Messvariablen (EMG/Kinematik) in den beschriebenen Belastungssituationen (Heben/Gang mit Perturbation) nachgewiesen werden. Gesunde Erwachsene zeigen bei einer kontinuierlichen Belastung des Rumpfes mit unterschiedlichen Gewichten (Heben von Lasten) keine Veränderung der Rumpfbewegung. Die plötzliche Belastung des Rumpfes durch Hinzunahme von Perturbationen der unteren Extremitäten im Gang konnte dagegen bei gesunden Probanden eine messbare Auslenkung des Rumpfes auf kinematischer als auch neuromuskulärer Ebene hervorrufen. Rückenschmerzpatienten zeigten ein verändertes neuromuskuläres und kinematisches Kompensationsmuster bei externen Perturbationen im Gang im Vergleich zu Gesunden. Dies ist charakterisiert durch eine verzögerte Reaktionszeit sowie verlängerte Dauer zum Erreichen der maximalen neuromuskulären Aktivität in Kombination mit einer reduzierten Lateralflexion des Rumpfes während Perturbation.

Die gewählte Testsituation - Perturbation im Gang - scheint, trotz Applikation der Perturbation über die Extremitäten und fehlender Fixierung des Beckens, geeignet zur Provokation erhöhter Anforderungen an die Stabilisierung des Rumpfes bei erwachsenen Probanden. Das veränderte neuromuskuläre und kinematische Kompensationsmuster bei Rückenschmerzpatienten kann als Surrogat einer erhöhten Belastung sowie einer reduzierten Rumpfstabilität bei Patienten bewertet werden. Das neu entwickelte Verfahren zur Erfassung der Rumpfstabilität ist geeignet, um Defizite bei Rückenschmerzpatienten zu identifizieren und folglich individuelle Zuordnungen zu Trainingsmaßnahmen vorzunehmen. Der Fokus in der Prävention bzw. Therapie von Rückenschmerzen sollte dem Folgend auf der Stabilisierung des Rumpfes mit dem Ziel der Verbesserung der neuromuskulären Kontrolle des Rumpfes in dynamischen Situationen liegen. Ein sensomotorischer Trainingsansatz (SMT) zur Optimierung der Rumpfstabilität und Kompensationsfähigkeit von unerwarteten externen Lasten ist zu präferieren.

1. Introduction

An essential function of the trunk is the compensation of external forces and loads in order to guarantee stability, as well as performance during everyday tasks and in high performance sports (Borghuis et al. 2008; Cresswell et al. 1994; Hodges et al. 2001; Kibler et al. 2006). Trunk stability is advantageous protecting against strain or injury during repetitive and suddenly applied loads (Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006). Hence, diminished trunk stability is discussed as a risk factor for the development of back pain, falls in older adults, injuries of the lower limb, and reduced performance capacity (Cholewicki et al. 2000; Grabiner et al. 2008; Zazulak et al. 2007). In addition, back pain (BP) has a high burden on the health systems of western industrial nations considering a lifetime prevalence of about 85 %, including high rates of chronification and physical restrictions (Balagué et al. 2012; Choi et al. 2010; Mannion et al. 2012; Mazaheri et al. 2013). There is still a matter of debate how to define trunk stability. The ability to maintain "trunk balance" despite of the presence of external mechanical forces or “neuro-muscular failure” characterises trunk stability. Therefore, the active compensation of external forces might play a major role in the prevention of injuries and overload. As a result, extraordinary relevance is ascribed to the optimisation or preservation of stability in situations with repetitive, highly dynamic, unexpected loading (Reeves et al. 2007; Reeves & Cholewicki 2009; Reeves et al. 2011). Here, the complex sensorimotor control and neuromuscular effectiveness are discussed as basic factors involved in adequate trunk stability (Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006; Reeves et al. 2011). Recent studies have analysed different loading situations on unstable surfaces implementing additional perturbations (Bazrgari et al. 2009; Eriksson Crommert & Thorstensson 2009; Milosevic et al. 2016; Radebold et al. 2000). Likewise, the EMG analysis of the trunk muscles reflected differences in the response times of healthy (H) and back pain patients (BPP) (Cholewicki et al. 2000; Radebold et al. 2000). Consequently, postural control as well as the neuromuscular activity of the trunk muscles are significant factors in the prediction of positive adaptations in situations with perturbations. Furthermore, analysis of the trunk motion is discussed beneficial to evaluate the mechanical factors associated with the development of back pain (Asgari et al. 2015; Gombatto et al. 2015; Maaswinkel et al. 2016; Müller et al. 2016a; Vogt et al. 2001;). Hence, there is an evident need to develop research-based strategies for the prevention and rehabilitation of back pain. Therefore, investigation of trunk function and stability comparing healthy subjects and back pain patients in dynamic movements is of primary interest to define adequate intervention regimes. In addition, a measurement protocol assessing three-dimensional motion and neuromuscular reflex activity in mainly dynamic, everyday life situations has not yet been validated.

Hence, the aims of the present thesis are:

(1) the development and definition of a reliable and valid measurement set-up to assess trunk motion and neuromuscular activity,

(2) the quantification of trunk motion and neuromuscular response as markers of trunk loading during everyday-life, sudden and continuous loading situations applied through the upper or lower limbs including perturbations, and

(3) the analysis of the (three-dimensional) motion and neuromuscular response as markers of stability in healthy and back pain patients in everyday loading situations.

In general, the present thesis is based on the following peer reviewed scientific articles:

1. **Müller J**, Engel T, Müller S, Kopinski S, Baur H, Mayer F. Neuromuscular response of the trunk to sudden gait disturbances: forward vs. backward perturbation. J Electromyogr Kinesiol. 2016;30:168–176. doi:10.1016/j.jelekin.2016.07.005 (Objective 1)

2. **Müller J,** Müller S, Stoll J, Rector M, Baur H, Mayer F.: Influence of load on 3-D segmental trunk kinematics in one-handed lifting: A Pliot Study. J Appl Biomech. 2016 Oct;32(5):520-5. doi: 10.1123/jab.2015-0227. (Objective 2)

3. **Müller J**, Müller S, Engel T, Reschke A, Baur H, Mayer F. Stumbling reactions during perturbed walking: Neuromuscular reflex activity and3-D kinematics of thetrunk – A pilot study. J Biomech. 2016;49(6):933-8. doi: 10.1016/j.jbiomech.2015.09.041. (Objective 2)

4. **Müller J,** Engel T, Müller S, Stoll J, Baur H, Mayer F. Effects of sudden walking perturbations on neuromuscular reflex activity and three-dimensional motion of the trunk in healthy controls and back pain patients. (under review Plos One 2016). (Objective 3)

2. Theoretical background: Analysis of trunk stability and back pain

2.1 Concept of stability: Functional stability of the human joints

Stability of the human joints plays a principal role in understanding the human musculoskeletal system with its strain and loading capacity (Bruhn & Gollhofer 1998; Gruber et al. 2006; Riemann et al. 2002; Riemann & Lephart 2002a; 2002b). Nevertheless, an absolute definition of the stability concept is not possible due to the multidimensional nature of the human locomotor system (Reeves & Cholewicki 2009; Reeves et al. 2011). Stability can be described as a state that remains unchanged under static conditions, even if managed by external forces that would change this state without compensation of the musculoskeletal system. It is described as the ability to regain an originally stable state after an external perturbation has occurred. Riemann et al. (2002) define (joint) stability as the ability to carry out an adequate adaptation to external forces and to remain in a steady state or to return as quickly as possible to it. Further, functional joint stability is considered the ability to control and/or maintain functional joint homeostasis in spite of physical activity and motion (Riemann & Lephart 2002a; 2002b). In this context, research interest has previously focused on the lower limbs, especially isolating the knee and ankle joints (Bruhn & Gollhofer 1998; Bruhn 1999; Lubowitz et al. 2007; Melnyk et al. 2007; Melnyk et al. 2008; Gauchard et al. 2010; Riemann & Lephart 2002a; 2002b; Zazulak et al. 2006; Zazulak et al. 2007). The application and transferability of this acquired knowledge to other joints of the human musculoskeletal system is limited due to the different functional demands on each of them (Wu et al. 2002).

Components contributing to joint stability

Both active and passive stabilisation has been described as two essential components of functional joint stability (Baur et al. 2010a; Bruhn 1999; Riemann & Lephart 2002a; 2002b). Stability is guaranteed by the active structures of the sensorimotor system. Mechanical stabilisation of a joint is induced by the passive structures (e.g. bone, cartilage, ligaments, sinews). The collaboration of both components creates this functional joint stability, with the aim of active compensation of external loads, forces and/or perturbations while maintaining the physiological stability of the affected joint e.g. avoidance of structure and tissue damages.

Functions of joint stability

Adequate joint stability is beneficial in stabilising the joint structures during repetitive, unexpected sudden loading and protecting against complaints or injuries (Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006). Hence, reduced stability is discussed as a risk factor for the development of injuries and decreased performance efficiency (Cholewicki et al. 2000; Granata & England 2006;

Granata & Wilson 2001; Marras et al. 2005; Riemann & Lephart 2002a; 2002b). In this context, particularly ruptures of the anterior cruciate ligament and acute damages in the ankle joint have been well investigated. Limited sensorimotor abilities have been discussed, along with diminished joint stability of the knee or ankle joint (Gruber et al. 2006; Hewett 2005; Solomonow & Krogsgaard 2001). Zazulak et al. (2006; 2007) were able to prove that athletes, particularly females, with reduced neuromuscular trunk control show a greater risk of joint injury, especially at the knee.

2.2 Trunk stability

2.2.1 Definition of trunk stability

In the investigation of trunk stability, different terms are used synonymously in the international literature e.g. spine stability, core stability, trunk stability, and lumbar stability (Granata et al. 2004; Maaswinkel et al. 2016; McGill 2001; McGill et al. 2003; Reeves et al. 2007; Reeves & Cholewicki 2009; Reeves et al. 2011; Wirth et al. 2016). Nevertheless, most of the concepts define stability as the ability to maintain balance of the trunk despite the presence of mechanical perturbations or neuromuscular control failure (Granata & England 2006; Reeves et al. 2007; Reeves & Cholewicki 2009; Reeves et al. 2011). Trunk strength capacity and sensorimotor control are usually described as contributing components (Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006; Leetun et al. 2004; Peate et al. 2007). Hence, there is strong evidence that strength capacity is an important factor for stability and is therefore part of the compensation of sudden, unexpected loading (Baur et al. 2010b; Brown et al. 2003; Granata & Wilson 2001; Iwai et al. 2004; Lee et al. 1999; Müller et al. 2012; Wirth et al. 2016). In addition, a sensorimotor approach seems to be suitable within the paradigm of rapid compensation of external loading and perturbations.

Further investigations have proven that the subsystems of passive (e.g., intervertebral discs, bones) and active structures in combination with the muscles and the neural control system (reflex activity) are influencing factors of trunk stability (Brown et al. 2003; Granata et al. 2001; Granata & England 2006; Reeves et al. 2005; Reeves et al. 2007; Wu et al. 1998). Stability needs to guarantee the main functions of the spine, e.g., the compensation of strains, enabling motion as well as the avoidance of pain and injuries (Reeves et al. 2007). The main functions of the trunk muscles are to guarantee the stability of the trunk in very different movement situations (rotation / translation) which can lead to instability and to ensure controlled motions of the spine (Luoto et al. 1995). Besides this, an appropriately time-coordinated and selective activation of the muscles is necessary to maintain stability. McGill et al. (2003) describe that an inappropriate contraction of even one of the muscles involved (e.g., time-wise) might already lead to an interference of the stability. Therefore, neuromuscular control is essential to maintain dynamic balance of the individual spine segments as a whole despite varying strains, mechanical disorders and/or neuromuscular control failure (Granata &

England 2006). Besides this, inadequate neuromuscular activation patterns can provoke instabilities and lead to the development of back pain over the long term (Hibbs et al. 2008). However, it is known that the magnitude of muscle activity needed to ensure stability depends on the task (McGill et al. 2003; Wirth et al. 2016). Stability during daily life activities requires a low activity level of the trunk flexors and extensors (McGill et al. 2003; Kavcic et al. 2004). A correlation between back pain and altered neuromuscular response of the dorsal muscles (extensors) is evident. This is characterized by an increased response time (Radebold et al. 2000; Taimela et al. 1993), an inexpedient voluntary muscle activation pattern (Henry et al. 2006; Radebold et al. 2000; 2001;) and inadequate anticipation (Oddsson & De Luca 2003) during expected and unexpected loading. However, it is unknown whether people who better control their trunk muscles and react faster to instable situations avoid excessive overloading.

2.2.2 Assessing trunk stability

2.2.2.1 Continuous vs. sudden trunk loading

The evaluation of trunk loading and stability requires a suitable measurement set-up including functional testing (Maaswinkel et al. 2016). Previous and current studies have focussed on two main tasks. Sudden, unexpected loading with isolation of the trunk (e.g. quick-release experiments) (Brown et al. 2003; Maaswinkel et al. 2016; Radebold et al. 2000; Shahvarpour et al. 2014)(Fig. 1) and continuous, expected loading by the limbs (e.g. lifting of weights, walking) are used to simulate trunk loading (Burgess et al. 2009; Granata & Wilson 2001; Wu et al. 1998).

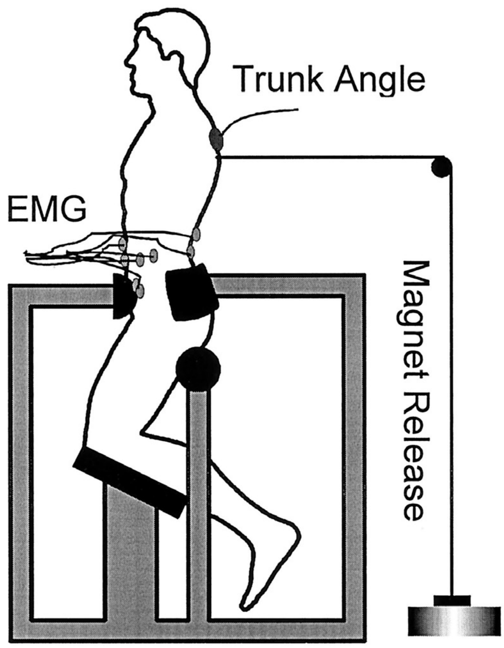

Figure 1: (Static) Quick-release experiment while load applied directly to the trunk (Radebold et al. 2001)

Sudden loading

In order to investigate compensation strategies to sudden loading, most of the studies applied loads directly to the trunk in initially static situations while standing or sitting (Cresswell et al. 1994; Granata et al. 2001; Maaswinkel et al. 2016; Moseley et al. 2002; Radebold et al. 2000; Vera-Garcia et al. 2007)(Fig. 1). Known as “sudden force release” or “quick-release” experiments, these aimed to isolate the trunk's response to external perturbations and to minimize the influence of distal body segments (Maaswinkel et al. 2016). Differences between healthy and BPP could be shown with regard to trunk response (e.g. neuromuscular reaction time) (Cholewicki et al. 2000; Radebold et al. 2000; 2001). Moreover, quick-release tests require sustaining a predefined static position. This usually did not correspond with daylike movements at the workplace or sport. Dynamic testing situations require the maintainance of the desired movement without task failure. This includes completely different demands to functional trunk stability. Therefore, transferability to sudden loading response in dynamic and thus more functional situations could be discussed critically. During daily life and sport activities, the trunk does not act isolated as was assumed in the quick-release experiments. The trunk always has to compensate sudden force while walking, running or jumping (Hibbs et al. 2008; Kibler et al. 2006). An everyday example of an unexpected, sudden loading situation is the compensation involved in slipping or tripping situations (Cholewicki et al. 2000; Cordero et al. 2003; Ferber et al. 2002; Grabiner et al. 1993; Granacher et al. 2010). Previous studies investigated lower leg perturbations while walking to analyse compensation strategies including neuromuscular activity and kinematic motion patterns (Dietz et al. 1987; 2004; Ferber et al. 2002; Keck et al. 1998; Oliveira et al. 2012; Tang et al. 1998; Taube et al. 2007). Nevertheless, it remains unclear whether a perturbation of the lower limbs leads to a notable displacement of the trunk.

Continuous loading

Functional and everyday situations include continuous loading generated by the limbs. In the current literature, various lifting tasks are often described to assess trunk stability (Burgess et al. 2009; Chow et al. 2004; Faber et al. 2007; Graham et al. 2011; Granata et al. 1999; Kingma & van Dieën 2004; Marras et al. 1999; McGill et al. 2013; van Dieën et al. 2003; Watanabe et al. 2012). Several studies investigated the influence of different weights (5 kg to 30 kg) as well as carrying conditions (two-handed vs. one-handed) on low back load (Burgess et al. 2009; Faber et al. 2007; Kingma & van Dieën 2004; McGill et al. 2013). It was concluded that one-handed carrying leads to higher low back loads compared to two-handed carrying of the same weight. Higher shear stress on the spine is discussed as a possible reason. Furthermore, lifting higher weights resulted in higher low back loads. As a conclusion, one-handed lifting could be seen as a more challenging situation that might be able to provoke different motion patterns of the trunk while using different weights. Since lifting tasks are omnipresent in daily life and correspond with expected large ranges of motion (ROM) of the trunk in multiple directions, they seem to be expedient in assessing trunk motion and neuromuscular activity (McGill et al. 2013).

The current state of the art in assessing trunk stability focuses on testing situations that include perturbations in combination with analysis of the neuromuscular activity as well as the kinematic movement pattern of the trunk (Maaswinkel et al. 2016). In a recent systematic review, Maaswinkel et al. (2016) postulate that perturbations should be applied unexpectedly and suddenly directly to the trunk. Pelvis and/or extremity movement should be isolated; otherwise the amount of the perturbation really addressing the trunk cannot be quantified. Nevertheless, it has to be questioned whether these isolated tests of the trunk are transferable to real-life movements where the limbs are always involved during loading and compensation of external forces (e.g. walking, lifting, jumping etc.).

In conclusion, assessment of trunk stability should focus on daily activities with normal to high-loading intensities, e.g. lifting or walking (Burgess et al. 2009; Granata & Wilson 2001; McGill et al. 2013; Wu et al. 1998;) Moreover, sudden (unexpected) as well as continuous (expected) loading should be implemented in the measurement protocol (Cholewicki et al. 2000; Maaswinkel et al. 2016; McGill et al. 2013). The implementation of perturbations during an automated human movement pattern could be used as an example of sudden loading. Therefore, stumbling while walking might be a reasonable testing situation. In addition, lifting various weights and normal walking might be a useful representation of continuous trunk loading.

2.2.2.2 Assessment of electromyography (EMG) and three-dimensional kinematics of the trunk

The active compensation of repetitive trunk loading with and without perturbations to optimize trunk stability has been described as highly relevant in the prevention and therapy of back pain (Hibbs et al. 2008; McGill 2010). In current studies, trunk stability was measured and quantified mostly by the use of individual (isolated) parameters (e.g., isokinetic trunk strength, neuromuscular activity (EMG)). The maximum strength capacity of the trunk is present as a decisive factor for stability and is therefore accepted as a relevant factor in the compensation of external loads and perturbations (Behm et al. 2010; Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006; McGill 2010; Prieske et al. 2015). In addition, Vera-Garcia et al. (2007) explain that both passive and active structures are essential for stabilization during performance, particularly while being perturbed. Nevertheless, it has not been finally clarified how neuromuscular abilities and the resulting trunk movement contribute to trunk stability besides the isolated strength capacity.

Therefore, the assessment of trunk muscle activity (EMG) should serve to clarify whether there is a correlation between neuromuscular control and trunk stability in dynamic loading situations (Dupeyron et al. 2010; Maaswinkel et al. 2016; Radebold et al. 2000). McGill et al. (2003) determine that trunk stability is the result of high-quality muscle coordination patterns that enclose several muscles and require a constant adaptation of the recruitment patterns as a function of the movement task. Effective stabilisation of the trunk is mainly achieved through task-dependent co-activation, as well as adequate recruitment and time patterns of the single muscles (McGill et al. 2003; Vera-Garcia et al. 2007). In the past, a 12-lead EMG set-up was used, which included six ventral (Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right sides) and six dorsal muscles (Mm erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of left and right sides) (McGill et al. 2003; Radebold et al. 2000; 2013; Zedka et al. 1998) (Fig. 2). In addition, analysis of the deep abdominal and back muscles (e.g., Mm. transversus abdominis, multifidus) is discussed relevant for the examination of stability (Vera-Garcia et al. 2007). The identification of one main stabilizing trunk muscle is however not reasonable due to the interaction of the ventral and dorsal muscles, as well as the dependence of the neuromuscular activity level on the loading task (McGill et al. 2003; Wirth et al. 2016). With the aim of evaluating trunk stability in highly dynamic strain situations, an application of the described 12-lead EMG set-up seems therefore feasible.

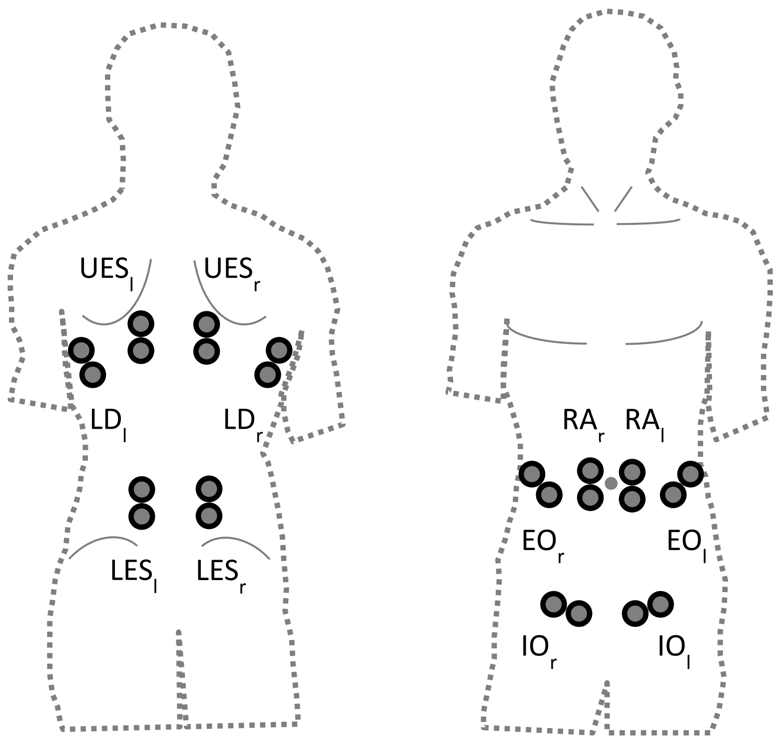

Figure 2: 12-lead EMG set-up of the trunk (derived from: Radebold et al. 2001)

Single muscles: $RA_{ri/le}$= M. rec. abd. right/left, $EO_{ri/le}$= M. obl. ext. abd. right/left, $IO_{ri/le}$= M. obl. int. abd. right/left; $LD_{ri/le}$= M. latis. dorsi right/left, $UES_{ri/le}$= M. erec. spinae thoracic (T9) right/left, $LES_{ri/le}$= M. erec. spinae lumbar (L3) right/left

To gain additional information about possible motion patterns of the trunk, kinematics are measured by means of (three-dimensional) motion analysis (Burgess et al. 2009; Joyce et al. 2012; Kingma & van Dieën 2004; Marras et al. 2005; Müller et al. 2015; Preuss & Popovic 2010; Schinkel-Ivy et al. 2015; Schinkel-Ivy & Drake 2015; Shahvarpour et al. 2014; 2015; Shin & Mirka 2004). This is considered necessary to evaluate the mechanical output of the trunk generated by the active and passive structures. However, differences in trunk motion depending on age, gender and back pain could already be shown in different static and dynamic tasks (e.g., lifting, walking) (Burgess et al. 2009; Gimmon et al. 2015; Müller et al. 2015; Seay et al. 2011a; Troke et al. 2005). In the past, often one rigid segment was used to assess trunk motion. However, these models neglect the intersegmental differences between lumbar and/or thoracic areas. Nevertheless, these differences could be relevant (Preuss & Popovic 2010; Schinkel-Ivy & Drake 2015). Recent studies have shown the need for multi-segmental kinematic trunk models in the evaluation of typical motion patterns during different loading tasks (Burgess et al. 2009; Leardini et al. 2011; McGill et al. 2013). Preuss & Popovich (2010) analysed the influence of the (kinematic) segmentation of the trunk on motion in healthy persons (sitting task). They compared three kinematic models with (1) only one rigid segment, (2) three segments and (3) seven segments. The authors concluded that the selection of the kinematic model significantly influences the motion measured during multi-planar movement tasks. In addition, they showed that a multi-segment model allows the differentiation of the motion patterns between specific spine areas (e.g., lumbar, thoracic). However, based on the movement tasks and the missing coverage of whole trunk by the applied model, its transferability to more functional and dynamic everyday tasks is questionable (e.g., lifting loads). Consequently, a reasonable analysis of trunk stability should include the assessment of the motion and activity of the trunk muscles.

2.3 Trunk stability and back pain

Chronic non-specific (low) back pain with a lifetime prevalence of 85 %, high rates of chronification and the resulting physical restrictions place a large burden on the societies and health systems of western industrialised nations. These are connected with high direct (e.g., therapy measures) and indirect costs (e.g., loss of working hours) (Balagué et al. 2012; Choi et al. 2010; Mannion et al. 2012; Mazaheri et al. 2013). Hence, there is an evident need to develop research-based strategies for the prevention and rehabilitation of back pain. Therefore, the investigation of differences in trunk function and stability between healthy and back pain patients is of primary interest in order to define adequate intervention regimes.

In the etiology, repetitive micro-trauma and insufficiency of the muscle-tendon complex based on inadequate postural and neuromuscular control, reduced maximum trunk strength capacity and trunk muscle fatigue during dynamic loading have been discussed (Bono 2004; Trainor & Trainor 2004). Trunk stability is advantageous to protect the spine while sudden, unexpected and repetitive loading (Borghuis et al. 2008; Hibbs et al. 2008; Kibler et al. 2006;). Hence, reduced trunk stability is discussed as a risk factor for developing, recurring or ongoing back pain (Borghuis et al. 2008; Choi et al. 2010; Leetun et al. 2004). It is evident that maximum trunk strength capacity and neuromuscular control are considered important factors. Diminished strength capacity of the trunk muscles (extensors/flexors) could be shown in back pain patients in comparison to healthy subjects (Gruther et al. 2009; Lindsay & Horton 2006; Müller et al. 2012; Yahia et al. 2011;). Nevertheless, maximum trunk strength capacity is no adequate representation in certain (e.g., isokinetic) test situations for the compensation of single or repetitive loading events (including perturbation) in people with and without back pain. Hence, the development of more functional loading situations is required. To analyse trunk compensation strategies as response to external strains and perturbations in healthy and back pain patients, neuromuscular activity and coordination of the trunk muscles is often analysed (McGill et al. 2013; Radebold et al. 2000). In addition, the 3-dimensional motion of the trunk is investigated in different situations (Asgari et al. 2015; Marras et al. 2005; Müller et al. 2016; Steele et al. 2013) in order to evaluate differences between symptomatic and healthy individuals.

2.3.1 Effects of back pain on trunk muscle activity

Altered neuromuscular activity of the trunk muscles is evident in back pain patients (Callaghan 2010; Cholewicki et al. 2000; Hodges & Moseley 2003; Ferguson et al. 2004; Nelson-Wong & Hanada 2011; Nelson-Wong et al. 2012; Radebold et al. 2000; 2001;).

With the help of "quick-release" experiments, several investigations were able to detect longer response times by the trunk muscles as a (stabilizing) response to unexpected, suddenly applied perturbations in back pain patients (Cholewicki et al. 2000; Gregory et al. 2008; Radebold et al. 2000). Further, Radebold et al. (2001) exemplarily showed an average difference in the response times of approximately 15 % (10 ms) between healthy subjects and patients to a sudden, external load applied directly to the trunk in a (half-)seated position. In addition, patients showed increased variability in their muscular response times. Nevertheless, comparable evidence is still lacking for dynamic situations with additional, unexpected perturbation. Besides this, the authors also observed a higher co-contraction of the agonistic and antagonistic trunk muscles in BPP as response to the experimental task (Radebold et al. 2000). Healthy participants showed phasic neuromuscular reaction patterns of the agonists and antagonists without co-contraction phases. In addition, Nelson-Wong & Callaghan (2010) showed increased co-contractions of the trunk muscles in subjects developing back pain after 2 hours of prolonged standing. Therefore, Nelson-Wong & Callaghan (2010) concluded that co-contraction is a predictor for the development of back pain due to the already altered EMG activation without the presence of subjective pain in these affected subjects at the beginning of the measurement protocol. In addition, Nelson-Wong et al. (2012) were able to prove altered recruitment patterns of the trunk muscles (combined with the hip musculature) in back pain patients. Moreover, they showed that during the extension phase of a trunk flexion-extension task, back pain developers showed a premature activation of the lumbar extensors in relation to the activation of the M. gluteus maximus. Healthy controls activated the M. gluteus maximus first, and the lumbar extensors only at the end of the extension phase. For these observations, comparable proof of more dynamic tasks is still pending in quasi-static situations.

The described increased co-contractions, altered recruitment patterns and extended response times in back pain patients are seen in relation to reduced trunk stability and are considered as predictors for the development of back pain (Nelson-Wong & Callaghan 2010; Nelson-Wong et al. 2012; Radebold et al. 2000; 2001). As a result higher loading of the spine as well as a reduced range of motion in the trunk are discussed (Marras et al. 2005).

2.3.2 Effects of back pain on trunk kinematics

A correlation between trunk motion and the risk of developing back pain is evident (Ferguson et al. 2004; Granata and England 2006; Marras et al. 2005; McGill et al. 2013). Besides this, reduced motion amplitudes as well as increased motion variability are evident kinematic alterations of the trunk in back pain patients during tasks (lifting; walking) with repetitive or continuous loading (Granata & England 2006; Larivière et al. 2000; Marras et al. 2005). Lariviere et al. (2000) examined the motion of the trunk in the sagittal and frontal planes in healthy subjects and BP patients during maximum trunk flexion. The authors were not able to identify any differences in the maximum angles of the trunk between healthy and back pain patients except for (right-sided) lateral flexion during additional loading (Larivière et al. 2000). Marras et al. (2005) analysed significant differences in trunk motion between healthy and back pain patients during different (symmetrical / asymmetrical) two-handed lifting tasks. Back pain patients showed a significantly lower motion speed and acceleration of the trunk. Nevertheless, Lariviere et al. (2000) and Marras et al. (2005) analysed trunk motion using two-dimensional measurements or EMG-based biomechanical modelling. No segmental analysis of the trunk motion was carried out.

Analyses of motion patterns and variability in trunk kinematics during gait are controversially in back pain patients (Lamoth et al. 2006; Müller et al. 2015; Seay et al. 2011a; Vogt et al. 2001;). Vogt et al. (2001) showed an increased stride-to-stride variability in lumbar trunk motion in all planes in BPP. Differences in the ROM as well as in the motion patterns between healthy and BP patients could not be proven. In addition, Steel et al. (2014) showed that back pain patients have increased motion variability only in the sagittal and transverse planes during normal walking compared to healthy controls. Furthermore, some studies report that symptomatic participants show a reduced rotation of the lumbar spine (Gombatto et al. 2015), while others demonstrated an increased rotation of the spine or pelvis (Seay et al. 2011b; 2011a). In this context, Müller et al. (Müller et al. 2015) described a reduced pelvis but comparable rotation motion of the trunk in BP patients compared to healthy controls during normal walking on different surfaces. To differentiate between persons with and without BP, the analysis of the motion in the transverse plane is of more importance than that in the sagittal plane (Müller et al. 2015).

3. Research objectives

The overall objective of the present thesis is the combined analysis of the 3D trunk kinematics and the neuromuscular (reflex) response to sudden loading in healthy subjects and back pain patients. Therefore, a measurement set-up to assess trunk motion and neuromuscular activity needs to be developed and validated. Moreover, trunk motion und neuromuscular response to sudden and continuous trunk loading in highly dynamic, everyday activities initiated by the upper or lower limbs is quantified. The outcome measurements are determined by means of the experimental approach and the defined loading situations.

For this thesis, the following research questions are defined:

F1: Can (3D) trunk motion and neuromuscular activity be reliably and validly measured during continuous and sudden loading?

F2: How do highly dynamic sudden and continuous loadings affect 3D kinematics and neuromuscular control of the trunk during everyday tasks?

F2$_a$: Are there differences in 3D trunk kinematics during continuous trunk loading while lifting objects with different weights?

F2$_b$: Does sudden loading (perturbation) while walking affect 3D trunk kinematics and neuromuscular control in healthy individuals?

F3: How does sudden loading affect trunk stability (quantified by 3D motion and neuromuscular activity) in healthy subjects and back pain patients?

The following main hypotheses can be derived:

H1: (3D) Trunk motion and neuromuscular activity can reliably be measured by means of a newly developed multi-segmental kinematic marker set-up in combination with a 12-lead EMG set-up in continuous and sudden loading situations.

H2$_a$: Increased weights lead to an altered 3D kinematic pattern during one-handed lifting.

H2$_b$: The use of additional perturbations during normal gait leads to alterations in 3D kinematics and neuromuscular reflex activity pattern compared to normal gait in healthy individuals.

H3: Back pain patients show altered trunk stability quantified by increased EMG latency and increased trunk ROM and motion variability during sudden loading by perturbations while walking.

4. Methodological approach of the thesis

The methodological construction of the thesis orientates itself, following the main objectives, around three phases: (1) examining the reliability of the newly developed measurement set-up to quantify dynamic trunk motion, (2) investigating the effect of additional continuous and suddenly applied loads on three-dimensional motion and neuromuscular control of the trunk in everyday tasks (lifting / walking with and without perturbation), and (3) examining the effect of suddenly applied loads on trunk stability (quantified by the three-dimensional motion and neuromuscular control) in healthy and back pain patients.

The structure of the project is outlined below (Fig. 3). The structure influences the selection of participants included and the methods used for the three different parts:

Project phase 1 (question F1)

The phase 1 focused on examining the reliability of a functional kinematic measurement set-up including neuromuscular activity assessment in a test-retest design (3 separate studies). Besides this, the influence and validity of different loading intensities (e.g., use of different perturbation stimuli during stumbling) on typical outcome measurements of trunk stability (e.g., EMG amplitudes) was analysed. (Paper 1)

Project phase 2 (question $F2_{a/b}$)

In phase 2, the effects of continuous and suddenly applied load on the three-dimensional motion and neuromuscular (reflex) activity of the trunk (cross-sectional study designs) were analysed. Validation of the defined measurement protocol (lifting / stumbling) was conducted to assess trunk stability in different loading situations in healthy participants. (Paper 2 / Paper 3)

Project phase 3 (question F3)

In the final phase, the influence of suddenly applied loads on trunk stability was analysed in a comparison of healthy and back pain patients. Therefore, the three-dimensional motion and neuromuscular activity of the trunk muscles were investigated with respect to highly dynamic loading in everyday situations (stumbling) for both groups. (Paper 4)

Accordingly, three reliability and three cross-sectional studies were conducted to define and validate the measurement set up. In addition, a fourth cross-sectional study was carried out with healthy and back pain patients analysing the effect of pain on trunk stability.

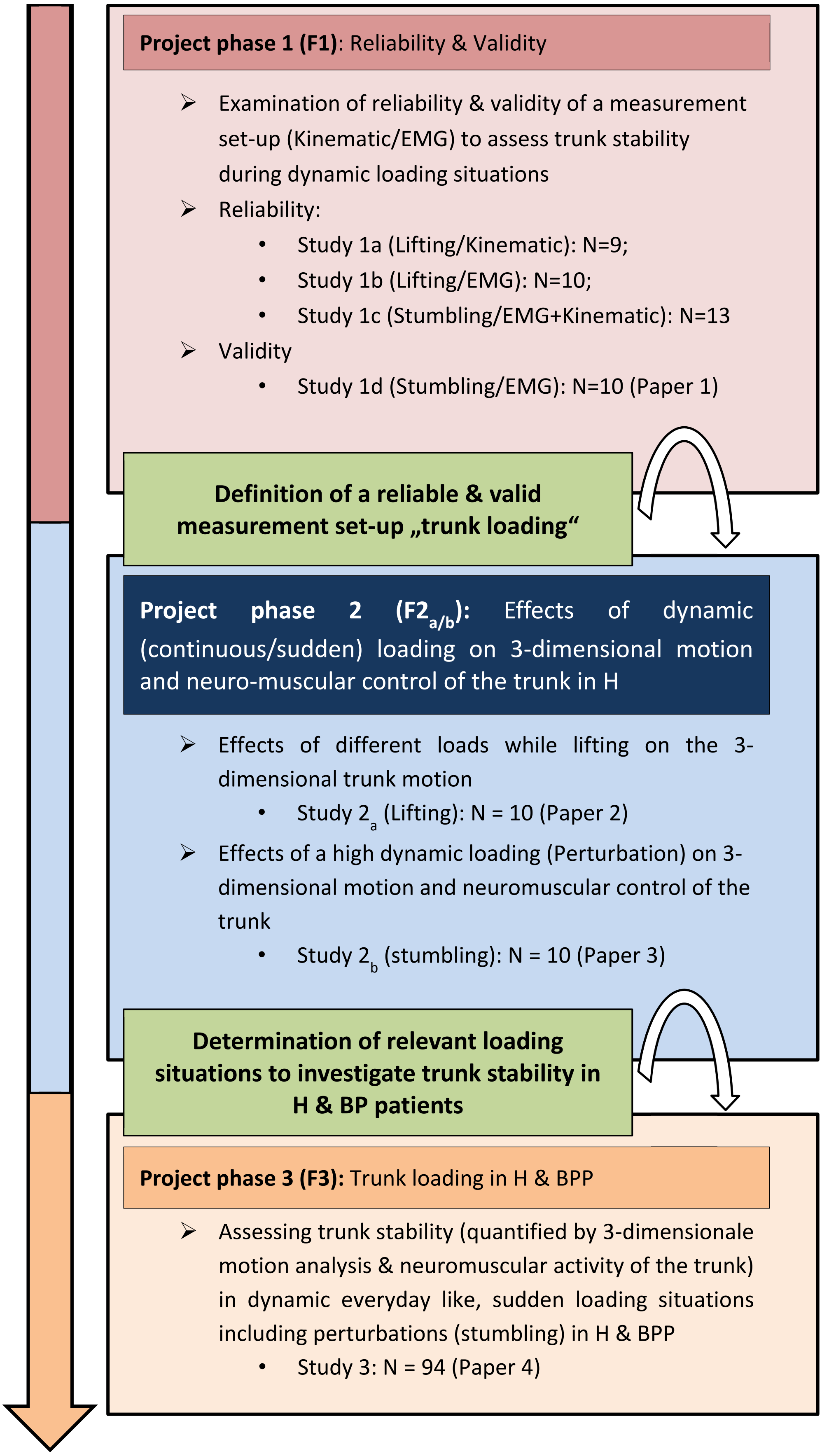

Figure 3: Project flow chart „Trunk loading and back pain“

Overall, the thesis is based on the four above listed scientific articles. Characteristics are detailed in Tab. 1.

Table 1: Characteristics of the original research papers included in the present thesis

Paper	Journal	Design	Participants	Measures	Chapter
Paper 1	J Electromyogr Kinesiol (peer-reviewed)	Cross-sectional	F: 5, M: 8 Mean age: 28±3 yrs	Stumbling, EMG activity/latency	5.2
Paper 2	J Appl Biomech (peer-reviewed)	Cross-sectional	F: 4, M: 6 Mean age: 29±3 yrs	Lifting, Trunk kinematics	6
Paper 3	J Biomech (peer-reviewed)	Cross-sectional	F: 5, M: 5 Mean age: 29±3 yrs	Stumbling, Trunk kinematics, EMG activity	7
Paper 4	PLOS One (under review)	Cross-sectional	F: 57 M: 37 Mean age: 29±9 yrs H: 80; BPP: 14	Stumbling, Trunk kinematics, EMG activity/latency	8

F = female; M = male; H= healthy controls; BPP = back pain patients

5. Can (3D) trunk motion and neuromuscular activity be reliably and validly measured during continuous and sudden loading?

5.1 Reliability

5.1.1 Background

To gain additional information about the motion pattern of the trunk in everyday strain situations, the motion analysis (kinematics) is pulled up beside the analysis of trunk muscle activity (Burgess et al. 2009; Fernandes et al. 2016; Leardini et al. 2011; Müller et al. 2015). Further, in previous studies one rigid segment was used to describe the motion of the trunk in different static or dynamic tasks (Kingma & van Dieën 2004; McGill et al. 1999). Nevertheless, these models neglect possible and relevant segmental differences between the thoracic or lumbar areas of the torso that could concern the development of back pain and/or injury.
Recent studies have shown the need for multi-segmental models in evaluating trunk motion characteristics in different strain situations (Burgess et al. 2009; Leardini et al. 2011; Preuss & Popovic 2010; Schinkel-Ivy and Drake 2015). Preuss & Popovich (2010) examined the influence of segmentation on the trunk motion using three different kinematic models consisting of different segments. The authors concluded that a multi-segment model permitted the differentiation of individual motion patterns (Preuss & Popovic 2010). Based on the movement task – sitting – transference to more functional, dynamic everyday tasks has to be questioned. Moreover, the developed model does not cover the whole trunk and therefore is not able to evaluate particularly the transitional areas between the thoracic and lumbar region. Besides this, all models used have in common that only little is known about their reliability (McGill et al. 1999; Preuss & Popovic 2010).
Therefore, the purpose of the studies was to evaluate the reliability of a novel and newly developed kinematic trunk marker set-up, with a focus on functional and anatomically defined segments, to validly assess trunk motion in everyday tasks (e.g., lifting of loads; walking with and without perturbations). In addition, the reliability of a valid 12-lead EMG set-up in static tests (Radebold et al. 2000; Zedka et al. 1998) aiming to assess neuromuscular activity of the trunk should be evaluated in the used dynamic tasks. A total of three studies, which are presented in Chapters 5.1.2 – 5.1.5, were carried out.

5.1.2 Reliability Study I: (Three-dimensional) trunk motion during the lifting of light and heavy loads

Nine participants (5 m/4 w; 29±3 y, 179±9 cm; 74±15 kg), all free of complaints and without a history of back pain, were included in the study. The study was carried out in a test-retest design with 7 days between both identical measurement days (M1 / M2). The measuring protocol always began with an anthropometric assessment (e.g., height, weight, age). Next, subjects were prepared for trunk motion analyses (see kinematics). Afterwards, a 5-minute non-specific warm-up on a bicycle ergometer was completed. The main task, lifting different loads, followed. Specifically, all participants had to lift a light (1 kg; L) and a heavy (10 kg; H) weight (one-handed; left) from the ground (position: left side of forefoot) onto a table (3 repetitions each condition; table height: 0.75 m, grab height: 0.19 m above the ground) in random order. Subjects were positioned in a hip-wide bipedal stance. The weights were lifted by slightly bending the knees and the trunk at a self-selected, moderate speed. The movement task was demonstrated by the investigator and every subject received a familiarization trial for each weight.

Kinematics: Trunk motion was assessed by means of a three-dimensional motion analysis system (Vicon; 8 cameras; MX3; 200 Hz). Therefore, 12 reflective markers were positioned over bony structures of the spine, framing three functional segments (upper thoracic segment (upper thoracic area [UTA]); lower thoracic segment (lower thoracic area [LTA]) and lumbar segment (lumbar area [LA])). The reflective markers were positioned with bilateral adhesive tape around the spinous processes of T3, T6, T9, T12, L3, and S1, as well as 0.05 m to the left and right of the spinous processes of C7, T6 and T12.

The maximum range of motion amplitudes (ROM; [°]) in extension / flexion (EF), lateral flexion (LF) and rotation (RO) were calculated for all 3 segments for L and H. The statistical analysis was done descriptively (mean ± standard deviation). With regards to the reproducibility analysis, the following parameters were calculated as recommended by Atkinson (Atkinson & Nevill 1998) and Bland & Altman (Bland & Altman 1986): ICC [2.1]: Intraclass Correlation Coefficient (Shrout & Fleiss 1979), "BIAS" and “Limits of Agreement” by Bland & Altman (Bland & Altman 1986), TRV: Test-Retest variability (König et al. 2012).

The highest ROMs were observed in M1 and M2 for all 3 segments for anterior flexion. The lowest ROM was calculated for the lateral flexion in all segments, being the smallest for UTA. The ICC varied from 0.66 - 0.99 for L and 0.11 - 0.86 for H. In all 3 segments and on two planes (EF; RO), the TRV was below 15% with an average of 10.5 ± 3.0 %. Only in lateral flexion for all 3 segments were higher TRV values observed. The BIAS (mean ± SD) was on average 0.17 ± 2.87 ° for L and -0.15 ± 4.48 ° for H. Results are detailed for all outcomes in Table 2.

Table 2: Absolute ROM values (M1 / M2) and indicators of reliability (ICC; TRV; BIAS; LoA) for a. light (L) and b. heavy (H) lifting for all segments on all planes

A. light (L) 1kg

Segment	Ebene	M1 [°]		M2 [°]		ICC [a.u.]	TRV [%]		BIAS [°]	LoA [°]	
		mean	SD	mean	SD		mean	SD		+	-
UTA	EF	104.41	15.53	104.00	10.79	0.76	8.25	3.84	0.40	18.16	-17.35
	LF	-0.65	19.88	-2.21	21.20	0.99	14.94	12.23	1.55	7.39	-4.29
	Ro	-56.15	14.57	-56.05	13.35	0.95	7.67	3.11	-0.09	8.95	-9.14
LTA	EF	112.17	18.28	117.69	20.12	0.90	6.30	4.47	-5.53	8.20	-19.25
	LF	24.06	9.76	20.10	6.31	0.69	29.97	10.78	3.96	14.67	-6.74
	Ro	-33.29	10.97	-36.68	12.98	0.77	13.95	14.17	3.39	17.88	-11.10
LA	EF	86.03	15.68	88.30	12.47	0.86	7.48	4.06	-2.27	11.64	-16.19
	LF	22.72	7.55	22.14	9.66	0.86	16.57	9.53	0.57	9.32	-8.18
	Ro	-30.23	8.32	-29.76	6.99	0.82	11.79	8.60	-0.47	8.41	-9.35

B. heavy (H) 10kg

Segment	Ebene	M1 [°]		M2 [°]		ICC [a.u.]	TRV [%]		BIAS [°]	LoA [°]	
		mean	SD	mean	SD		mean	SD		+	-
UTA	EF	102.90	13.37	101.83	6.83	0.11	9.74	8.03	1.07	27.26	-25.11
	LF	-11.16	28.44	-11.47	21.08	0.82	52.22	20.09	0.31	29.01	-28.39
	Ro	-55.66	10.35	-57.30	7.66	0.69	11.83	7.29	1.65	15.10	-11.80
LTA	EF	107.61	18.73	117.87	19.25	0.86	9.23	3.39	-10.26	-2.90	-17.62
	LF	26.82	9.26	20.72	6.23	0.57	26.01	18.00	6.10	16.97	-4.76
	Ro	-35.39	9.20	-37.27	8.32	0.73	15.45	13.12	1.88	13.95	-10.19
LA	EF	82.88	7.82	86.06	7.31	0.26	8.52	6.24	-3.18	13.69	-20.04
	LF	25.62	7.26	25.12	7.41	0.79	14.53	9.68	0.51	9.74	-8.73
	Ro	-29.88	6.56	-30.45	6.55	0.82	11.82	8.22	0.57	8.03	-6.90

SD= standard deviation
ICC= Intraclass Correlations Coeffizient; TRV = Test-Retest-Variability; LoA= Limits of Agreement

5.1.3 Reliability Study II: neuromuscular activity of the trunk muscles during lifting

Ten participants were enrolled in the study (6 m/4 w; 30±4 y, 178±8 cm; 72±12 kg). The study was conducted in a test-retest design (7 days between the two measurement days (M1 / M2)). The measuring protocol began with an assessment of the anthropometrics (e.g., height, weight, age). First, all subjects were prepared for investigation of trunk neuromuscular activity (see EMG). Next, two-handed lifting of a heavy load (20 kg) up onto a table in front of the subjects followed. This test served as a reference for the EMG normalization of the one-handed lifting task. The major task, one-handed (left-sided) lifting of a heavy load, followed this one. All participants lifted a heavy weight (20 kg; H) from the ground onto a table (3 repetitions; table height: 0.75 m, grab height: 0.19 m above the ground). Subjects stood in a hip-wide bipedal stance, starting the lifting task by slightly bending the knees and the trunk at a self-selected, moderate speed. The lifting task was demonstrated by the investigator, and every subject received a familiarization trial before the testing conditions were applied (two-handed / one-handed).

EMG: The neuromuscular activity of the trunk was measured by means of a telemetrical surface EMG (RFTD-32, myon AG, Baar, Switzerland). Twelve pairs of EMG electrodes (Radebold et al. 2000) were positioned over six ventral (Mm rec. abd. (RA), obl. ext. abd. (EO), int. abd. (IO) on the left and right) and 6 dorsal muscles (Mm erec. spinae thoracic (UES) /lumbar (LES), latis. dorsi (LD) on the left and right). The EMG amplitudes (RMS; %) were normalized to two-handed lifting (reference task with 20 kg) and analyzed for the entire lifting cycle. In addition, the mean (normalised) EMG-RMS for 4 trunk areas (McGill (2013)) was calculated: ventral right ((VR); RA, EO, IO of the right side); ventral left ((VL); RA, EO, IO of the left side); dorsal right ((DR); UES, LES, LD of the right side) and dorsal left ((DL); UES, LES, LD of the left side). Statistical analysis was done descriptively (mean ± SD) followed by calculation of the reliability indicators as recommended by Atkinson (Atkinson & Nevill 1998) and Bland & Altman (Bland & Altman 1986): ICC [2.1]: Intraclass Correlation Coefficient (Shrout & Fleiss 1979), "BIAS" and “Limits of Agreement” by Bland & Altman (Bland & Altman 1986), TRV: Test-Retest variability (König et al. 2012).

The highest EMG-RMS was observed in the RA of the right side (M1/M2), and the UES showed the lowest EMG-RMS on the left side. Considering the individual muscles, the ICC ranged between -0.13 and 0.90 (Table 3). The TRV was on average 35.1 ± 13.6 %, but only for the UES (left-side) could values below 20 % be calculated. The BIAS ranged between -87 and 20 %. The detailed results are presented in Table 3 for the individual muscles and Table 4 for the muscle groups.

The highest level of activity in M1 and M2 is shown for the right ventral muscle group, and the lowest for the dorsal muscle group of the left side (Table 4). Regarding the reliability of the 4 trunk areas, the ICC ranged between 0.13 and 0.75 and the mean TRV was 28.6 ± 5.7 %.

Table 3: Absolute values (standardised EMG-RMS) and reliability indicators for the neuromuscular activity of the trunk muscles during lifting (one-handed; left-sided)

Muskeln	M1 [%]		M2 [%]		ICC [a.u.]	TRV [%]		BIAS [°]	LoA [%]	
	mean	SD	mean	SD		mean	SD		+	-
M. rectus abdominis ri	164	59	153	43	0.72	22.00	10.83	11	83	-61
M. rectus abdominis le	132	32	119	39	0.62	20.97	14.39	13	70	-45
M. externus abdominis ri	530	310	703	316	0.63	44.93	33.47	-87	402	-575
M. externus abdominis le	151	69	107	35	0.30	38.63	26.73	44	156	-68
M. internus abdominis ri	112	131	74	57	0.04	59.89	41.90	-4	133	-141
M. internus abdominis le	152	67	172	94	0.67	26.87	24.79	-20	103	-143
M. latissimus dorsi ri	310	170	340	169	-0.13	54.99	34.04	-30	438	-499
M. latissimus dorsi le	198	116	173	102	0.62	39.87	26.40	25	202	-153
M. erector spinae (T9) ri	92	39	88	23	0.62	25.84	14.79	4	58	-49
M. erector spinae (T9) le	67	26	70	33	0.90	17.66	10.66	-3	21	-28
M. erector spinae (L3) ri	116	29	97	36	0.32	29.33	27.78	20	88	-49
M. erector spinae (L3) le	78	38	83	35	0.65	40.03	30.49	-5	53	-63

SD= standard deviation; ri = right; le = left
ICC= Intraclass Correlations Coeffizient; TRV = Test-Retest-Variability; LoA= Limits of Agreement

Table 4: Absolute values (standardised EMG-RMS) and reliability indicators for neuromuscular activity of the muscle groups of the trunk during lifting

Muscle groups	M1 [%]		M2 [%]		ICC [a.u.]	TRV [%]		BIAS [%]	LoA [%]	
	mean	SD	mean	SD		mean	SD		+	-
VR	241	100	272	127	0.58	29.05	21.51	-32	164	-228
VL	145	50	133	49	0.75	22.36	14.86	12	77	-53
DR	166	62	167	59	0.13	36.03	21.33	-1	148	-151
DL	118	55	110	46	0.74	27.07	19.91	8	78	-62

SD= standard deviation; ri = right; le = left
ICC= Intraclass Correlations Coeffizient; TRV = Test-Retest-Variability; LoA= Limits of Agreement

Legend:
VR = ventral muscles right (rec. abd., ext abd., int. abd. of right side)
VL = ventral muscles left (rec. abd. le, ext. abd., int. abd. of left side)
DR = dorsal muscles right (lat. dors.; erec. T9; erec. L3 of right side)
DL = dorsal muscles left (lat. dors.; erec. T9; erec. L3 of left side)

5.1.4 Reliability Study III: (Three-dimensional) motion and neuromuscular activity of the trunk during normal gait with and without perturbation

Thirteen participants (7 male / 6 female; 28 ± 3 y; 179 ± 10 cm; 76 ± 13 kg) were included in the study. The study was conducted featuring a test-retest design (7 days between the two measurement days (M1/M2)). After assessing the anthropometrics (height, weight, age), subjects were prepared for analysis of trunk motion (Vicon, UK; 8 cameras; 200 Hz) and neuromuscular activity (RFTD-32, myon AG, Baar, Switzerland). Therefore, twelve reflective markers were positioned over the bony structures of the spine, framing 3 functional segments: upper thoracic (upper thoracic area [UTA]), lower thoracic (lower thoracic area [LTA]) and lumbar segments (lumbar area [LA])) More specifically, markers were attached (with double-sided tape) to the spinous processes of T3, T6, T9, T12, L3, and S1, as well as 0.05 m to the left and right of the spinous processes of C7, T6 and TH2. Further, twelve pairs of EMG electrodes (Radebold et al. 2000) were located over six ventral (Mm rec. abd. (RA), obl. ext. abd. (EO), int. abd. (IO) on the left and right) and six dorsal muscles (Mm erec. spinae thoracic (UES) /lumbar (LES), latis. dorsi (LD) on the left and right). This was followed by a 5-min. familiarization walk on the split-belt treadmill (Woodway, Weil am Rhein, Germany) without perturbations. Next, all subjects walked (velocity: 1 m/s) on the split-belt treadmill while 5 right- and left-sided perturbations (treadmill belt decelerating, 40 m/s^2, 50 ms duration; 200 ms after heel contact) were randomly applied. After every perturbation, a break of at least 10 seconds guaranteed that a normal gait pattern was restored before the next perturbation. For safety reasons, all subjects wore a hip belt that was connected to an emergency stop release. Trunk motion and neuromuscular activity were measured during normal gait (G) and stumbling (gait with perturbations; S). The maximum motion amplitudes (ROM; [°] in extension / flexion (EF), lateral flexion (LF) and rotation (RO)) were calculated for all 3 segments. The EMG amplitudes (RMS; [%]) were evaluated for a time interval of 200ms after perturbation. In addition, the amplitudes were normalized to the identical time interval of the normal gait (average step from 5). The mean (normalised) EMG-RMS for 4 trunk areas according to McGill (2013) was calculated: ventral right (right ventral area ((VR); RA, EO, IO of the right side); ventral left (left ventral area ((VL); RA, EO, IO of the left side); dorsal right (right dorsal area ((DR.); UES, LES, LD of the right side) and dorsal left (left dorsal area ((DL); UES, LES, LD of the left side). Only right-sided perturbations were evaluated for data analysis. Statistical analysis (mean ± SD) was done descriptively followed by reliability analysis, including ICC [2.1]: Intraclass Correlation Coefficient (Shrout & Fleiss 1979), "BIAS" and “Limits of Agreement” to Bland & Altman (Bland & Altman 1986) and the TRV: Test-Retest variability (König et al. 2012).

Neuromuscular activity of the trunk muscles

EMG activity of all muscle groups was increased during stumbling, ranging from 4- to 7-fold activity on both days (M1/M2) compared to normal gait. The mean TRV for all 4 areas was 38.1 ± 2.26 %. The

ICC varied between 0.17 (DL) and 0.39 (VL) and averaged 0.31 ± 0.10. For normal gait (G), the ICC was 0.32 ± 0.49 on average and varied between -0.19 and 0.89. The TRV averaged 17.8 ± 5.7 % during G. The detailed results for all calculated outcome measurements are shown in Table 5 (ICC, TRV, BIAS, LoA).

(Three-dimensional) trunk kinematics

Trunk motion showed the highest absolute values for rotation of all segments during normal gait as well as stumbling (13.08 ° - 16.07 ° (Table 6), lateral flexion showed the smallest ROM (3.30 ° - 3.57 °). TRV averaged 22.1 ± 10.2 % during stumbling and 19.7 ± 12.0 % during stumbling. The ICC varied from 0.06 (LA in LF) to 0.92 (LTA in AF) during stumbling (S) and averaged 0.64 ± 0.32. During normal gait (G), ICC was on average 0.29 ± 0.63 and varied between -1.05 and 0.94. The TRV for G averaged 19.70 ± 11.99 %. All results are detailed in Table 6 for all outcome measurements.

Table 5: Reliability of the trunk's neuromuscular activity (EMG-RMS) during normal gait with (A) perturbation and (B) without perturbation

A

muscle group	M1 [%]		M2 [%]		ICC	TRV [%]		BIAS [%]	LoA [%]	
	mean	SD	mean	SD		mean	SD		+	-
VR	605	297	552	240	0.30	40.12	34.15	53.09	659.66	-553.48
VL	543	225	456	197	0.39	37.53	19.04	86.91	522.02	-348.19
DR	664	278	692	345	0.38	39.70	30.09	-28.27	637.26	-693.71
DL	441	193	472	92	0.17	35.15	29.91	-31.53	338.16	-401.23

B

muscle group	M1 [%]		M2 [%]		ICC	TRV [%]		BIAS [%]	LoA [%]	
	mean	SD	mean	SD		mean	SD		+	-
VR	102	14	105	11	0.89	12.49	8.11	-3.04	24.99	-31.06
VL	98	16	106	12	-0.19	18.84	13.74	-8.13	33.15	-49.41
DR	98	15	108	20	0.54	14.38	10.16	-10.53	18.45	-39.51
DL	95	19	110	22	0.03	25.27	17.10	-15.33	38.41	-69.06

SD= standard deviation; ICC= Intraclass Correlations Coeffizient; TRV = Test-Retest-Variability; LoA= Limits of Agreement

Legend:
VR = ventral muscles right (rec. abd., ext abd., int. abd. of right side)
VL = ventral muscles left (rec. abd. le, ext. abd., int. abd. of left side)
DR = dorsal muscles right (lat. dors.; erec. T9; erec. L3 of right side)
DL = dorsal muscles left (lat. dors.; erec. T9; erec. L3 of left side)

Table 6: Absolute ROM (M1 / M2) and reliability indicators (ICC; TRV; BIAS; LoA) for three-dimensional trunk motion on 3 planes (A) during perturbed gait and (B) normal gait

A. gait with perturbation

Segment	Plane	M1 [°]		M2 [°]		ICC	TRV [%]		BIAS [°]	LoA [°]	
		mean	SD	mean	SD		mean	SD		+	-
UTA	EF	7.35	2.45	7.47	3.38	0.85	23.97	18.77	-0.12	3.04	-3.29
	LF	9.40	4.72	7.55	2.01	0.37	40.94	16.88	1.85	9.29	-5.59
	Ro	13.73	4.66	14.27	4.36	0.73	15.41	13.50	-0.54	5.75	-6.83
LTA	EF	8.25	2.72	8.48	3.21	0.92	12.44	9.32	-0.23	1.83	-2.28
	LF	9.22	3.57	8.18	1.93	0.27	31.22	20.01	1.04	7.51	-5.43
	Ro	14.25	4.20	13.88	4.62	0.85	15.88	7.47	0.37	5.10	-4.36
LA	EF	7.04	2.78	7.12	2.70	0.80	17.79	19.71	-0.08	3.26	-3.42
	LF	7.65	2.22	7.41	1.89	0.06	30.34	14.43	0.23	5.52	-5.05
	Ro	16.07	4.61	15.16	4.11	0.88	10.59	10.95	0.92	4.75	-2.91

B. normal gait (without perturbation)

Segment	plane	M1 [°]		M2 [°]		ICC	TRV [%]		BIAS [°]	LoA [°]	
		mean	SD	mean	SD		mean	SD		+	-
UTA	EF	5.18	1.77	5.87	2.43	0.83	14.47	13.49	-0.69	1.40	-2.78
	LF	3.66	0.77	3.57	1.19	-0.03	31.23	23.67	0.09	2.61	-2.43
	Ro	12.69	1.11	13.40	2.04	0.55	7.34	7.91	-0.71	2.00	-3.43
LTA	EF	7.11	1.48	7.54	1.07	0.29	16.06	9.77	-0.43	2.32	-3.18
	LF	3.30	1.29	3.47	0.74	-1.05	42.89	27.51	-0.17	3.24	-3.58
	Ro	12.38	2.66	12.39	3.19	0.88	9.22	7.96	-0.01	2.72	-2.74
LA	EF	5.53	1.42	6.29	1.03	0.23	23.88	7.04	-0.76	1.93	-3.44
	LF	3.79	1.39	3.59	0.18	-0.03	24.20	16.40	0.20	2.68	-2.28
	Ro	14.45	3.98	14.00	3.67	0.94	7.97	4.66	0.45	2.79	-1.89

UTA: upper thoracic area; LTA: lower thoracic area; LA: lumbar area; EF: extension/flexion; LF: lateral flexion; RO: rotation.
SD= standard deviation; ICC= Intraclass Correlations Coeffizient; TRV = Test-Retest-Variability; LoA= Limits of Agreement

5.1.5 Discussion and Conclusion

Trunk motion (kinematics) during lifting of different loads and walking with / without perturbation

The results of the studies show acceptable to excellent reproducibility for trunk motion assessment (in all 3 segments / planes) in dynamic strain situations (lifting different loads; gait with/without perturbation). Similar or higher ICCs could be shown in comparison to previous investigations (Bible et al. 2010; Fernandes et al. 2016; Schinkel-Ivy et al. 2015; Van Daele et al. 2007). The test-retest variability, with the exception of lateral flexion (segment LTA/LA), was less than 15 % during lifting and less than 25 % during stumbling. Lateral flexion has already been described with reduced reliability in previous studies (Bible et al. 2010; Brink et al. 2011; Leardini et al. 2011; Schinkel-Ivy et al. 2015; Van Daele et al. 2007). Van Daele et al. (2007) report limited reliability of lateral flexion in the two-segment model (pelvis/trunk) during a seated movement task. Schinkel-Ivy et al. (2015) reported diminished reliability for movement tasks with low motion amplitude. This was true for all three segments (UTA, LTA, LA) in the lateral flexion of the trunk in the loading situations described above (one-handed lifting, gait with/without perturbation). According to Schinkel-Ivy et al. (2015), an increased number of repetitions for each task might be valid for increasing reliability measures. In comparisons of the two loading tasks examined (lifting/stumbling), it becomes obvious that lifting as well as the normal gait can be discussed as stable, automated movement patterns (Burgess et al. 2009; McGill et al. 2013). Therefore, reliability analysis of trunk motion measurements reveals values ranging all the way to excellent. In contrast, walking with perturbation reflects a highly variable loading scenario represented by reduced reliability. An increase in the number of repetitions might be considered suitable for the reduction of variability and for stabilisation of the reliability analysis.

Neuromuscular activity of the trunk during lifting of loads and walking with / without perturbation

The results reveal insufficient reproducibility values for the neuromuscular activity of the trunk muscles during lifting and stumbling. The high variability during lifting can be explained, on the one hand, by the high degree of freedom of the trunk (no fixation of body segments) during the motion task. The reliability of neuromuscular activity during normal walking (without perturbation) has to be described as acceptable to excellent. This was expected based on the automated movement pattern of the human gait (Dietz et al. 2002; Duysens & Van de Crommert 1998). Application of perturbations during normal walking was observed to increase the variability to ~45 %, particularly for the areas of the trunk flexors. Consequently, the reliability decreases as a result. Nevertheless, considering the extent of the perturbation evoked increase in the normalized EMG-RMS, this must be put into perspective (e.g., 40 % variability vs. 605 % increase (ventral right group)). Furthermore, the magnitude of the changed activation is deductive. However, it is unclear whether a higher number of repetitions (perturbations) could lead to a reduction in the variability, and therefore a stabilisation of the reaction pattern, as described by Schinkel-Ivy et al. (2015). This could also effect an improvement in the reliability (through a reduction in the variability).

5.2 Validity: Neuromuscular response of the trunk to sudden gait disturbances: forward vs. backward perturbation

Neuromuscular response of the trunk to sudden gait disturbances:
forward vs. backward perturbation

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Short title: Trunk response to perturbed walking

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Reference:
Müller J, Engel T, Müller S, Kopinski S, Baur H, Mayer F. Neuromuscular response of the trunk to sudden gait disturbances: forward vs. backward perturbation. J Electromyogr Kinesiol. 2016;30:168–176. doi:10.1016/j.jelekin.2016.07.005.

5.2.1 Abstract

The study aimed to analyze neuromuscular activity of the trunk comparing four different perturbations during gait.

Thirteen subjects (28±3yrs) walked (1m/s) on a split-belt treadmill, while 4 (belt) perturbations (F1,F2,B1,B2) were randomly applied. Perturbations differed, related to treadmill belt translation, in direction (forward (F)/backward (B)) and amplitude (20m/s² (1) /40m/s² (2)). Trunk muscle activity was assessed with a 12-lead-EMG. EMG-RMS [%] (0-200ms after perturbation; normalized to RMS of normal gait) was analyzed for muscles and four trunk areas (ventral left/right; dorsal left/right). Ratio of ventral:dorsal muscles were calculated. Muscle onset [ms] was determined. Data analysis was conducted descriptively, followed by ANOVA (post-hoc Tukey-Kramer (α=0.05)).

All perturbations lead to an increase in EMG-RMS (428±289%). F1 showed the lowest and F2 the highest increase for the flexors. B2 showed the highest increase for the extensors. Significant differences between perturbations could be observed for 6 muscles, as well as the 4 trunk areas. Ratio analysis revealed no significant differences (range 1.25 (B1) to 1.71 (F2) between stimuli. Muscle response time (ventral: 87.0±21.7ms; dorsal: 88.4±17.0ms) between stimuli was only significant (p=0.005) for the dorsal muscles.

Magnitude significantly influences neuromuscular trunk response patterns in healthy adults. Regardless of direction ventral muscles always revealed higher activity while compensating the walking perturbations.

Keywords: stumbling, gait perturbation, EMG, core, MiSpEx*

5.2.2 Introduction

Stability provided by the muscles encompassing the trunk is an important factor in determining trunk response to sudden, unexpected loading. Consequently, compensation of external loads ensures the stability and performance of the body during daily life as well as in dynamic high-intensity activities (Cresswell et al. 1994; Kibler et al. 2006; Hibbs et al. 2008). Unexpected, high-loading situations may occur after slipping or tripping to prevent falls (Grabiner et al. 1993; Magnusson et al. 1996; Cholewicki et al. 2000; Ferber et al. 2002; Cordero et al. 2003; Granacher et al. 2010; Shahvarpour et al. 2014). Experimentally induced stumbling during locomotion is often used to investigate the response of the body and its isolated segments where dynamic postural control is disturbed (Cordero et al. 2003; Granacher et al. 2010; Pijnappels et al. 2005; Tang et al. 1998). Previous studies have additionally shown that the muscle reflex activity of the trunk muscles might be partly responsible for the compensation of slipping events simulated during lower leg perturbations (Stanek et al. 2011; Tang et al. 1998;). Moreover, the involvement of the trunk as part of the compensation strategy of walking perturbations is evident (Grabiner et al 1993; Müller et al. 2016; van der Burg et al. 2005). In this respect, Müller et al. (2016) reported that healthy subjects revealed a specific neuromuscular pattern in response to a high-intensity backward perturbation. Increased abdominal activity as well as increased activity of the trunk muscles ipsilateral to the point of application of the perturbation could be observed. This was the case even though no statistically significant differences between muscles (abdominal vs. back; ipsilateral vs. collateral) could be found. Van der Burg et al. (2005) observed fast muscle onset (60-80ms) in trunk muscles with abdominal muscles representing significant faster onset compared to back muscles after a tripping perturbation. Authors concluded that shown trunk muscle response pattern aimed to reduce trunk forward bending decreasing the risk of falling (van der Burg et al. 2005). However, the experimental setup (repeated trials over a 1.5m long walkway) used did not allow the analysis of trunk response during continuous automated human walking.

In recent literature, influence of perturbation characteristic on trunk muscle response is analyzed mostly during sitting and standing (Jones et al. 2008; Shahvarpour et al. 2015; Torres-Ovideo & Ting 2007; Zedka et al. 1998). Zedka et al. (1998) concluded in this context that muscles respond with different intensities to different perturbation magnitudes. Studies on multi-directional postural disturbances during standing presented different neuronal compensation strategies with respect to the perturbation direction (Jones et al 2008; Torres-Oviedo & Ting 2007). However, the transferability of these results to dynamic situations with respect to functional trunk stability could be discussed critically. Oliveira et al. (2012) added that direction dependent neuromuscular adaptation strategies of the leg as well as the trunk muscles could be observed during multi-directional walking perturbations using a restricted walkway with moveable platforms.

Trunk stability, led by neuromuscular (reflex) activity, is required to control trunk motion during repetitive (dynamic) loading situations and protect the trunk of overloading. Importantly, reduced trunk stabilization, mainly characterized by an increased muscle response time, higher co-contraction and altered muscle activity pattern, could contribute to overloading and injury, e.g. back pain patients. As an example, inadequate neuromuscular control during dynamic loading is discussed as an explanatory model in back pain etiology (Bono 2004; Lawrence et al. 2006). In addition, during dynamic task the trunk is exposed to high loading and extra forces compared to static testing (e.g. sitting or standing). Therefore, stumbling while walking seems to be a suitable functional testing situation in assessment of trunk stability (Müller et al. 2016). No study has yet investigated the neuromuscular reflex activity of the trunk in a comparison of different perturbations during continuous (treadmill) walking. Therefore, it remains unclear whether different perturbations during an automated movement pattern (normal gait) lead to different responses of the trunk muscles as a relevant factor for load compensation strategies. Therefore, the purpose of this study was to analyse the neuromuscular reflex activity of the trunk muscles during four different walking perturbations. It is hypothesized that the direction as well as the magnitude of the perturbation will provoke different neuromuscular response patterns in the trunk muscles.

5.2.3 Material and methods

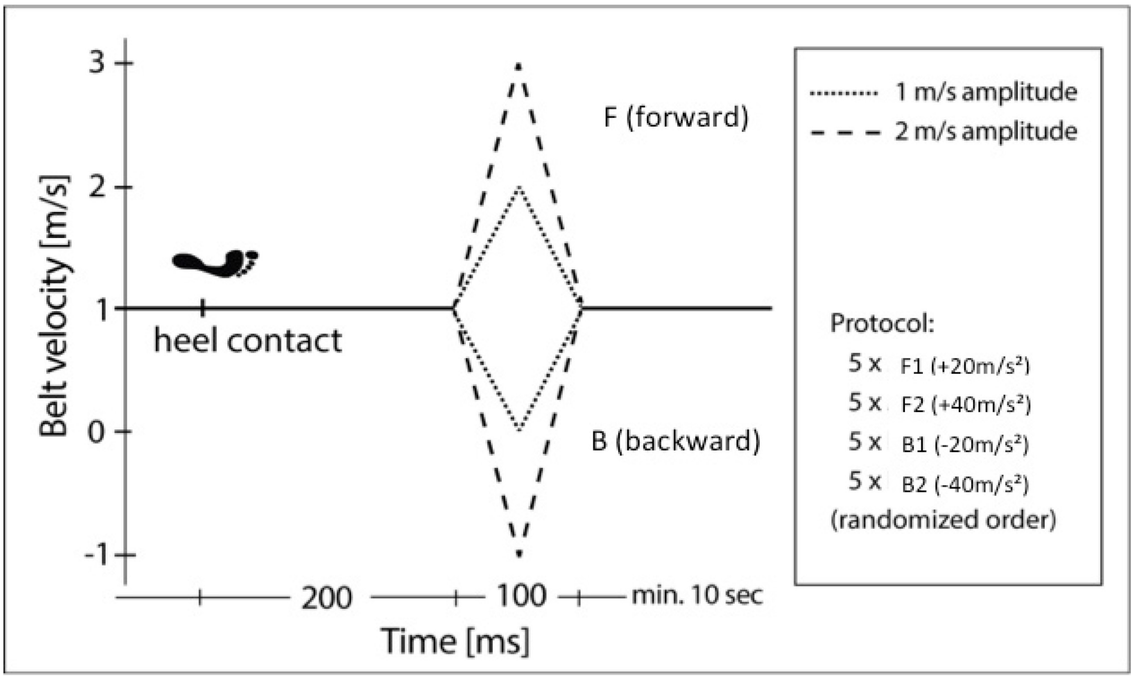
Subjects

13 adults (8 male / 5 female; 28 ± 3yrs; 180 ± 10cm; 77 ± 12kg) with a minimum of 2 hours of physical activity per week, free of complaints and with no history of back pain, were included in the study. Before voluntary participation, all subjects read and signed their written informed consent. The University´s Ethics Commission approved the study.

Measurement protocol

After assessment of the anthropometrics (height, weight), subjects were prepared for surface electromyographic measurements (EMG) of the trunk. For this, 12 pairs of EMG electrodes were positioned over ventral and dorsal muscles (Radebold et al. 2000; Zedka et al. 1998;). Afterwards, a short warm-up including a familiarization walk on the split belt treadmill (Woodway, Weil am Rhein, D) without perturbations (5mins at 1 m/s) followed (Winter & Yack 1987). Thereafter, each subject walked for about 10 minutes at a baseline velocity of 1m/s while 5 repetitions of four different right- and left-sided stimuli were randomly applied. The stimuli differed in direction (acceleration/deceleration) and amplitude (2-fold vs. 3-fold) (Fig. 4A). Overall, subjects were commanded to walk as normal as possible on the treadmill while randomly perturbations will be applied. As a consequence, subjects walked on the treadmill while knowing that perturbations will be applied but not knowing when (time), where (leg) and how (direction). In addition, participants were instructed to compensate the stimuli aiming to get back to normal upright walking pattern within the following three to four steps (Cordero et al., 2003). For safety reasons, all participants were secured by a waist belt connected to an emergency stop release. (Fig. 4B)

A.



B.

C.

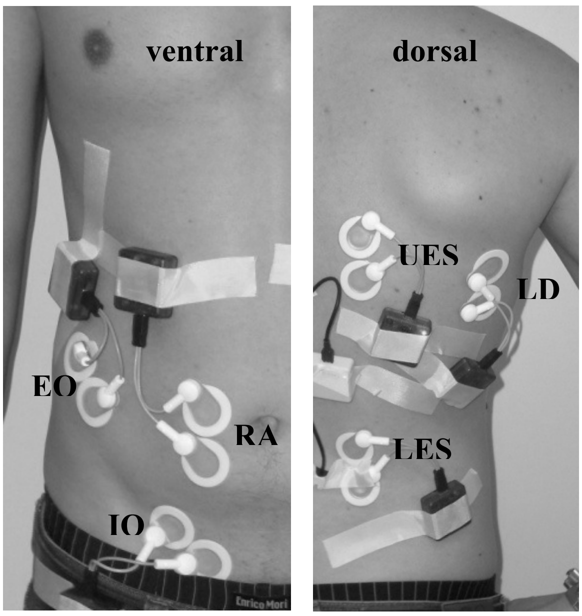


Figure 4: Technical and methodological set-up of the stumbling protocol

A. Perturbation characteristics for the 4 different stimuli

B. Customized Split belt treadmill with 2 separate selectable belts (left) with prepared Subject

C. 12-lead EMG-trunk-setup (RA: M rec. abd., EO: M. obl. ext. abd. IO: M. obl. int. abd; UES: M. erec. spinae thoracic (T9), LES: M. erec. spinae lumbar (L3), LD: M. latis. dorsi)

Stumbling characteristics

A computer controlled split-belt treadmill (Woodway, Germany; Fig. 4B), powered by two electric engines, was used to generate rapid perturbations (max. acceleration: 40 m/s^2) of velocity decreases independently for each belt. Baseline velocity and perturbation impulses (amplitude, duration; Fig. 4A) were controlled by a customized software (stimuli, pfitec, Endingen, Germany). The different stimuli were each applied 200ms after initial heel contact triggered by a plantar pressure insole (Pedar X, Novel, Munich, D). This ensures that participants are perturbed in the early phase of the gait cycle (weight acceptance) and single support phase bearing already full load of body weight on the foot (Dicharry 2010). The acceleration/deceleration phase lasted 50ms, with an additional 50ms until the return to baseline velocity. In relation to the treadmill belt movement, the stimuli were applied in either the forward (acceleration; F) or backward (deceleration; B) direction. Additionally, the stimuli amplitudes were increased one-fold by 20m/s^2 (1) or two-fold by 40m/s^2 (2). This resulted in the use of the following 4 stimuli: backward one-fold (B1): -20m/s^2 (stimuli velocity: 0m/s), backward two-fold (B2): -40m/s^2 (stimuli velocity: -1m/s), forward one-fold (F1): +20m/s^2 (stimuli velocity: +2m/s) and forward two-fold (F2): +40m/s^2 (stimuli velocity: +3m/s) randomly applied during walking (Fig. 1A). Forward one- and two-fold (F1/F2) aimed to simulate tripping, backward one- and two-fold (B1/B2) simulated slipping events. Results of reproducibility analysis indicated an excellent reliability of the chosen customized stumbling protocol. Test-retest-reliability (TRV) ranged between 3 to 7% for delay, duration and amplitude of the pre-set stimuli in gait. In addition, Bland & Altman analysis revealed a bias (±1.96*SD) between 0.0 (±16) ms and 5 (±37) ms for duration and delay (Engel et al. 2013).

Cordero et al. (2003) showed that more than one stride is necessary for recovery after gait perturbation. Consequently, a minimum of 10 seconds rest in-between two perturbations ensures that participants achieved their normal walking pattern after the previous perturbation and prior to the subsequent perturbation. To account for the triggering method used, only right-sided perturbations were applied in the final data analysis. Left-sided perturbations were randomly applied to ensure that participants did not adapt their normal walking pattern to only right-sided perturbations.

Electromyographic analysis

Muscular activity of the trunk was assessed using a 12-lead surface EMG during normal (unperturbed) gait and stumbling (Fig. 4C). Analysis included six ventral (Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right sides) and six dorsal (Mm erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of left and right sides) muscles (Fig. 1C) (Radebold et al. 2000; Zedka et al. 1998). Muscular activity was analyzed using bipolar surface EMG (band-pass filter: 5 Hz to 500 Hz, gain: 5.0, overall gain: 2500, sampling frequency: 4000 Hz; RFTD-32, myon AG, Baar, Schweiz). Localization of electrodes was carefully determined according to Radebold et al. (2000). Before electrodes (AMBU Medicotest, Denmark, Type N-00-S, inter-electrode distance: 2cm) were applied, the skin was shaved and slightly exfoliated to remove surface epithelial layers and finally disinfected. In addition, skin resistance was controlled (<5 kΩ). The longitudinal axes of the electrodes were in line with the presumed direction of the underlying muscle fibers. The signal was rectified before calculation of outcome measures. The first 5 unperturbed strides of the walking protocol were averaged to one mean unperturbed step. For each stimulus, 5 perturbed strides were averaged to one mean perturbed step. Root mean square (RMS; [%]) and neuromuscular response (onset; [ms]) analysis served as the main outcome measures. RMS were analyzed within the first 200 ms after perturbation to analyse the representative time window for neuromuscular reflex activity of the trunk's surrounding muscles (Granacher et al. 2010; Müller et al. 2016; Taube et al. 2007)(Fig. 5). The RMS was normalized to the RMS of the same interval during unperturbed walking (Granacher et al. 2010; Taube et al. 2007). Additionally, the mean (normalized) EMG-RMS of four trunk areas was calculated (right ventral area (VR; Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of right side); left ventral area (VL; Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left side); right dorsal area (DR; Mm erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of right side) and left dorsal area (DL; Mm erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of left side)) (McGill et al. 2013; Zedka et al. 1998). As a measure of co-contraction, ratio of ventral and dorsal muscles was calculated (formula: (mean RMS of all ventral muscles):(mean RMS of all dorsal muscles)).

In the time domain, the onset of muscular activity (ms) was detected and measured for the first active ventral and dorsal muscle, representing a response to the perturbation. In addition, the specific muscles (first on) for the ventral and dorsal muscles were documented. A manual detection method was used to define muscle activity onset conducted by one experienced investigator. Within this detection method, a visual detectable (continuous) increase in the averaged EMG signal was defined for onset detection. (Hodges & Bui 1996)

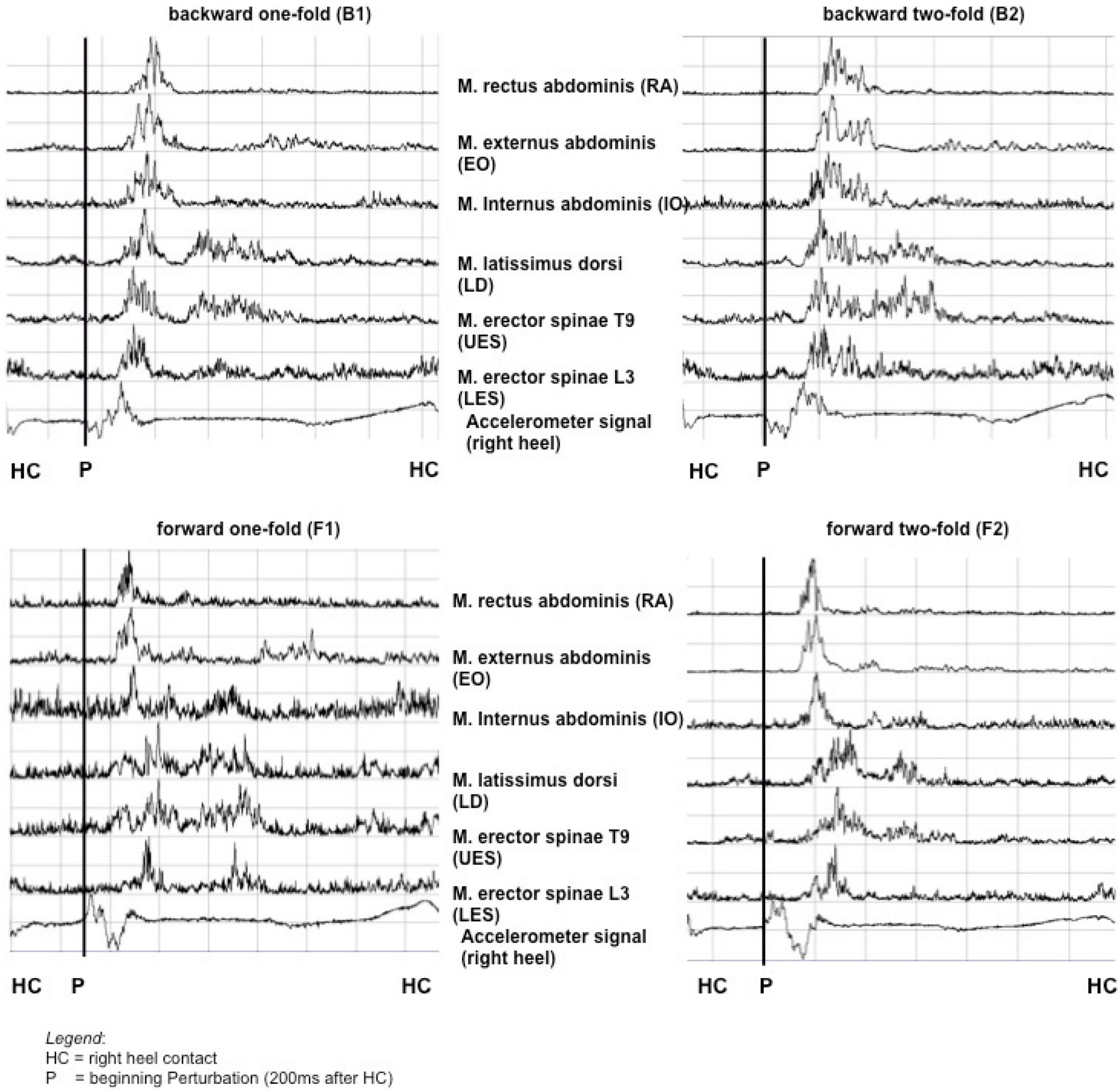


Figure 5: Neuromuscular response pattern of all six (right) trunk muscles during the four different stimuli (averaged, rectified signal of 5 (right-sided) perturbed steps for 1 subject)

Data analysis:

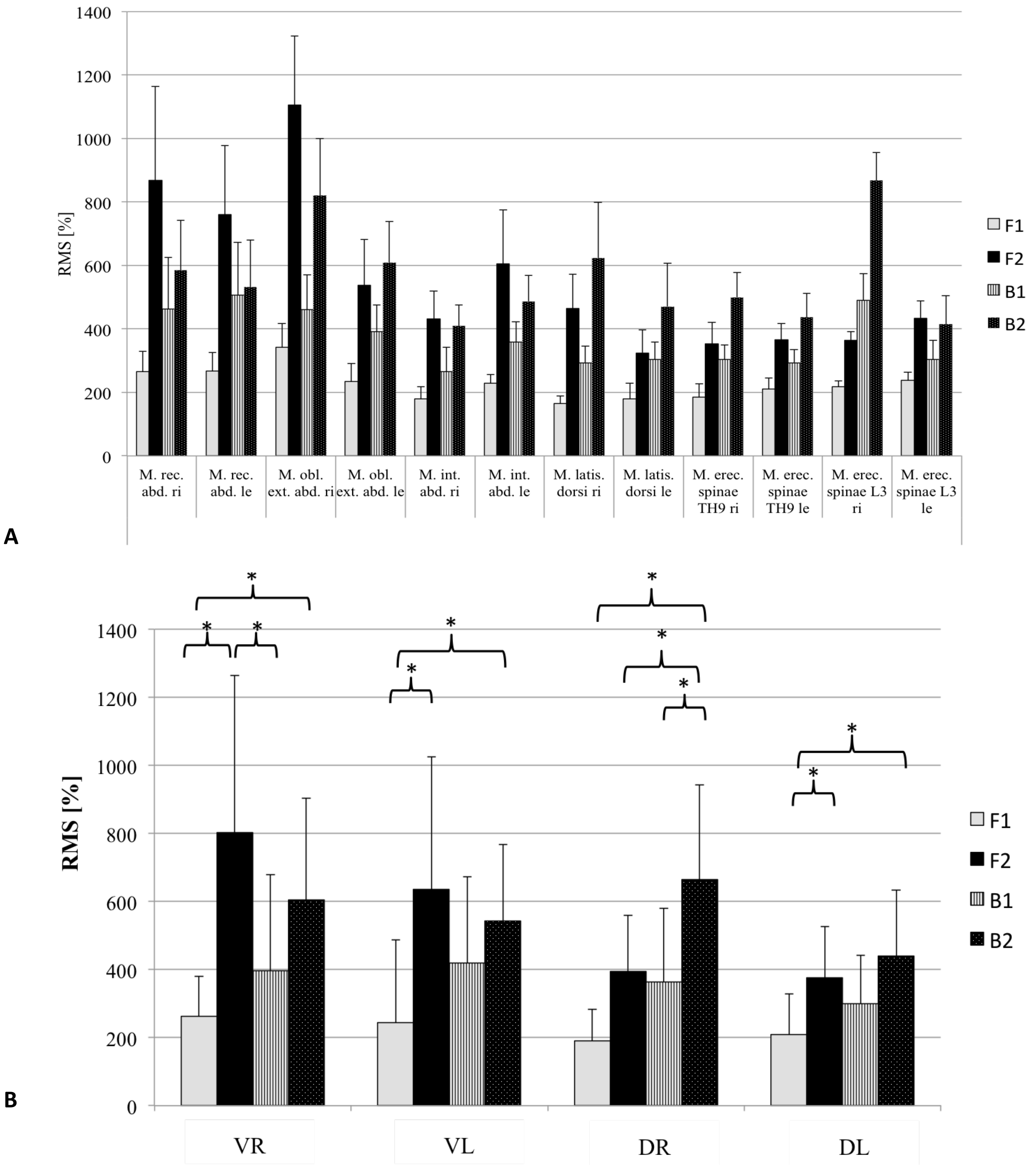
After a plausibility check, the data was descriptively analyzed, including the means and standard deviations (SD) of the EMG (RMS%; onset, [ms]) for normal gait and perturbed walking. One-way ANOVA (post hoc: Tukey-Kramer) was used to analyse RMS differences between perturbations for single muscles and trunk areas as well as ratio of ventral and dorsal muscles, with a Bonferroni adjustment for multiple comparisons. The level of significance was set at α=0.05 (adjusted: 12 muscles; α=0.004). In addition, one-way (comparison between 4 stimulis for ventral or dorsal muscles) and two-way ANOVA (factors: muscle (ventral/dorsal) * stimuli (F1/F2/B1/B2); interaction effect) were applied to the analysis of the muscle onset.

5.2.4 Results

Neuromuscular response pattern in comparison of the 4 different stimuli are visualized in Fig. 5. The analysis of the trunk muscle activation showed higher EMG amplitudes (RMS) for all ventral and dorsal muscles during stumbling compared to normal walking. This is valid for all four perturbations (Fig. 6A). Despite perturbation characteristics, the mean amplitude increase measured was 428±289% and ranged between 112 and 1852%. Results of the neuromuscular response patterns (RMS) of the trunk comparing the four different perturbations are detailed in Fig. 6 and 7. Stimuli forward one-fold (F1) showed the lowest EMG increase overall (range: 112-589%). Stimuli forward two-fold (F2) showed the highest increase for a ventral (EO right) and backward two-fold (B2) the highest for a dorsal (LES right) muscle (Fig. 6). The EMG-RMS for backward two-fold (B2) ranged between 170 and 1119%.

Statistically significant differences for EMG-RMS between perturbations could be observed for EO, LD, UES and LES. For the right-sided M. externus obliques (EO), significant differences between stimuli forward two-fold (F2) vs. backward one-fold (B1) and stimuli forward two-fold (F2) vs. forward one-fold (F1) ($p=0.0007$) could be observed. For the left-sided M. externus obliques (EO), significant differences between stimuli backward two-fold (B2) vs. forward one-fold (F1) and stimuli forward two-fold (F2) vs. forward one-fold (F1) were found ($p=0.003$). Right-sided M. latissimus dorsi (LD) showed differences between stimuli backward two-fold (B2) vs. backward one-fold (B1), backward two-fold (B2) vs. forward one-fold (F1) and forward two-fold (F2) vs. forward one-fold (F1) ($p<0.0001$). Right-sided M. erector spinae T9 (UES; B2 vs. F1; $p=0.001$) as well as M. erector spinae L3 (LES; B2 vs. F1; B2 vs. F2; $p=0.001$) also showed significant differences between the four perturbations. (Fig. 6A). Differences between perturbations were found in all 4 trunk areas ($p<0.004$)(Fig. 6B). Ratio of ventral and dorsal muscles revealed higher ventral activity, regardless of stimuli, and no statistical significant differences between the four stimuli applied ($p=0.13$). Ratio was 1.25 for stimuli backward one-fold (B1), 1.27 for stimuli backward two-fold (B2), 1.34 for stimuli forward one-fold (F1) and 1.71 for stimuli forward two-fold (F2).

Muscle onset analysis revealed an average onset time of 87.0 ± 21.7ms for ventral muscles and 88.4 ±17.0ms for dorsal muscles, regardless of stimuli applied. Over all four stimuli, in 51% of the cases EO left, in 27% EO right, in 10% IO right, in 10% IO left and in 2% RA left were the muscles showing first activity in the ventral muscles. For the dorsal muscles, in 45% UES right, in 31% LD left, in 18% UES left, and in 2% each for LD right, LES right/left. Differences between stimuli are detailed in Tab. 7 for both muscle groupes. In general, one-way ANOVA analysis showed statistical significant differences ($p=0.005$) for muscle onset between the 4 stimuli for the dorsal muscles (post hoc: F2 vs. B2, $p=0.0051$; F2 vs. B1, $p=0.016$) but not for the ventral muscles ($p>0.05$) (Fig. 8).



VR: ventral right (RA, EO, IO of right side); VL: ventral left (RA, EO, IO of left side);
DR: dorsal right (UES, LES, LD of right side); DL: dorsal left (UES, LES, LD of left side)
*significant differences (p<0.05)

Figure 6: EMG-RMS for the trunk muscles (normalized to unperturbed walking) as response to right-sided perturbations

A. EMG-RMS [%] for six ventral and six dorsal muscles

B. EMG-RMS [%] for 4 areas of the trunk

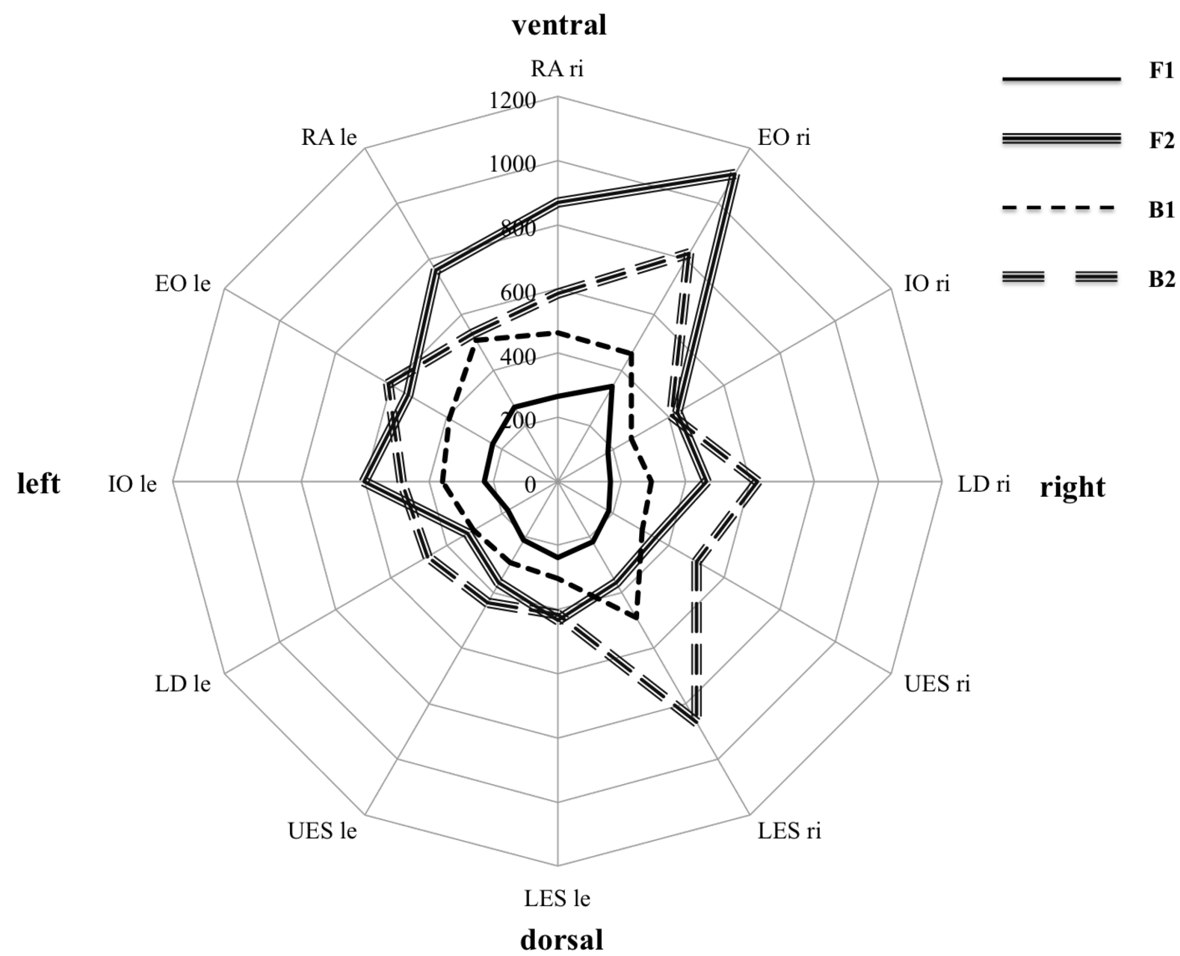


Figure 7: Polar plot showing mean EMG-RMS [%] for six ventral and six dorsal muscles for the 4 different stimuli as response to right-sided perturbations (ventral muscles: RA,EO, IO of left/right side; dorsal muscles: UES, LES, LD of left/right side)

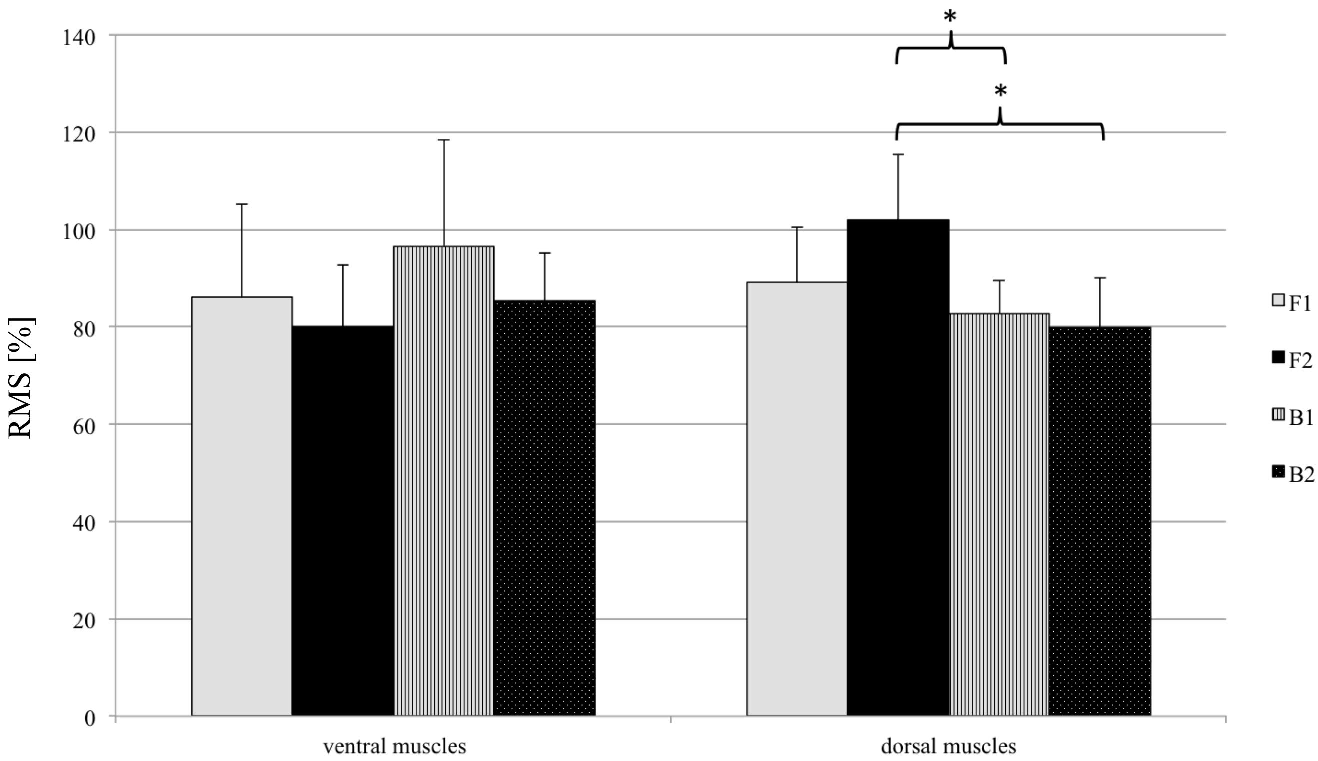


Figure 8: Neuromuscular trunk response (onset; [ms]) of first ventral and first dorsal muscle to the 4 different stimuli as response to right-sided perturbations (*significant differences (p<0.05))

Table 7: Muscle onset: Frequency [%] of first muscle onset for ventral and dorsal muscles as response to right-sided perturbations

	ventral muscles						dorsal muscles					
stimuli	**RA ri**	**RA le**	**EO ri**	**EO le**	**IO ri**	**IO le**	**LD ri**	**LD le**	**UES ri**	**UES le**	**LES ri**	**LES le**
F1	0	0	39	39	8	15	0	54	15	31	0	0
F2	0	0	42	42	8	8	0	46	46	0	9	0
B1	0	0	25	50	8	17	0	17	50	33	0	0
B2	0	8	0	75	17	0	8	8	67	8	0	8
all	**0**	**2**	**27**	**51**	**10**	**10**	**2**	**31**	**45**	**18**	**2**	**2**

5.2.5 Discussion

The purpose of this study was to analyze the neuromuscular response of the trunk's surrounding muscles to 4 different perturbations while walking. The main results show that provoked stumbling leads to increased trunk muscle activity (EMG-RMS) compared to normal walking, without consideration of perturbation characteristics. Furthermore, the magnitude of the stimuli significantly influences the level of neuromuscular response. Thus, regardless of direction (forward vs. backward) ventral muscles always revealed higher activity while compensating the applied perturbations. Muscle response time did only show significant differences between the 4 stimuli for the dorsal muscles.

The overall increased muscular activity of the trunk during gait perturbations is in line with literature (Stanek et al. 2011; Sakai et al. 2008; van der Burg et al. 2005), even though different outcome measures and methodological setups were used. Stanek et al. (2011) found up to 4-times higher EMG activity (average EMG; peak EMG) of the Mm. obl. ext. abd. during perturbed gait compared to normal walking. In this study, 3- (stimuli forward one-fold; F1) to 10- (stimuli forward two-fold; F2) times higher EMG activities of the Mm. obl. ext. abd. were analyzed. As already mentioned, methodological differences (perturbation characteristics; EMG analysis, normalization; outcome measures) hampered a more detailed comparison. Hence, these studies did not compare the influence of different stimuli on the neuromuscular reflex response of the trunk's surrounding muscles.

In line with the presented results, Zedka et al. (1998) showed by the use of the same EMG trunk muscle setup that, even in the sitting position, muscle response patterns of the trunk depended on the magnitude and direction of the perturbations applied. In contrast to the results presented here, the authors showed that forward perturbation during sitting leads to higher ventral activity and a backward perturbation to higher dorsal activity. In contrast, in a more dynamic situation such as stumbling while walking, the influence of the direction of perturbation via the lower limbs did not seem to have a relevant influence on the neuromuscular response pattern of the trunk muscles. From a methodological point of view, it must be mentioned that the definition of backwards and forwards was used conversely between the two studies due to the different reference systems (running split belt treadmill vs. static chair). This was considered for the interpretation and comparison of the results. In addition, Shahvarpour et al. (2015) showed increased muscle reflex activity of the trunk with increased loading during a forward perturbation (sitting) that concurs with the present results.

Taking all this into account, lower leg perturbations of greater magnitude should be used to evaluate the neuromuscular reflex activity pattern of the trunk muscles. In addition, ratio analysis showed that, regardless of stimuli used, ventral muscles show a higher neuromuscular reflex activity to ensure stability of the trunk while compensating walking perturbations of different directions and magnitudes. Therefore, ventral muscles contribute in a more distingt way to trunk stability while compensating walking perturbations (van der Burg et al. 2005).

Muscle response presented in our study ranged between 61 - 174ms for ventral, and 59 - 129ms for the dorsal muscles and generally agreed with findings of the literature where response latencies of the trunk muscles to surface perturbations varied between 70–250 ms while sitting, 100–200 ms while standing and 60 to 80ms while walking (Radebold et al. 2000; Milosevic et al. 2016; Preuss & Fung 2008; van der Burg et al. 2005; Zedka et al. 1998). The muscle response times to our applied walking perturbations can be classified as medium latency reflexes (Milosevic et al. 2016). Concluding that polysynaptic reflexes regulate the neuromuscular response to high-dynamic walking perturbations. Differences between the stimuli for the dorsal muscles might be discussed as a result of the high inter-individual variability of human reflex response.

A measure of displacement of the trunk (e.g. kinematics) is missing in the presented study and hampers a more detailed comparison of the severity of the perturbation between the four different stimuli. Nevertheless, the validity of the backward two-fold (B2) stimuli was presented elsewhere (Müller et al. 2016). These findings show that perturbation lead to a displacement of the trunk in all 3-segments and all planes analysed compared to normal walking (range 115 to 262% deviation of normal gait) but only being significant for lateral flexion. A transmission of those results to the other three stimuli (F1, F2, B1) remains speculative. However, Grabiner et al. (1993) observed a dominance of trunk motion in the sagittal plane during gait perturbation in the anterior direction discussing this as a decrease in postural stability, increasing the risk of falling. In relation to Grabiner et al. (1993), one could speculate that more dorsal muscle activity would not be beneficial and would instead decrease the stability of the body due to trunk extension and the elevation of its center of mass (Grabiner et al. 1993; Grabiner et al. 2008; van der Burg et al. 2005).

Perturbations are implemented with increasing frequency into prevention and rehabilitation training to enhance the level of difficulty of (sensorimotor) core stability exercises and/or to prepare subjects for compensation of unexpected sudden trunk loading, e.g. in back pain patients (Pedersen et al. 2004). Therefore, the transfer into training programs should include a diversity of perturbations (especially magnitude) due to the shown differences in neuromuscular patterns of the trunk muscles to different perturbations. Consequently, the trunk muscles close to the perturbation seem to play a more important role in compensating balance loss as fast as possible (McGill et al. 2013). Therefore, the point of application of the perturbation during balance exercises and targeting muscles should be located on the same side of the body (Müller et al. 2016; van der Burg et al. 2005).

Limitations

During stumbling, all subjects walked at the same baseline velocity, despite different anthropometrics and stride parameters. The influence of walking speed on the plantar pressure (Taylor et al. 2004), neuromuscular activity (Chumanov et al. 2007) and lower limb kinematics has been thoroughly investigated (Kang 2007). Nevertheless, for a standardized and comparable test situation between all subjects, a consistent walking velocity during the stumbling protocol was favored (Cordero et al. 2003). Nevertheless, it cannot be ruled out that, at this consistent velocity, subjects were stressed to different extents. In addition, no specific instructions regarding the task of the trunk during compensation was given to the participants. Use of different compensation strategies (e.g. leg-dominant, trunk-dominant) and amount of trunk involvement was not assessed. Different strategies might have influenced the neuromuscular trunk response pattern to different amount.

However, the transfer of the results presented is limited to the cohort analysed – middle-aged participants without back pain. If these findings hold in special cohorts, e.g. older adults, back pain patients or athletes, needs to be investigated.

In conclusion, provoked stumbling leads to an increase in EMG activity of the trunk muscles (polysynaptic reflex activity) compared to normal walking, despite different perturbation characteristics. Moreover, the magnitude of perturbation significantly influences the neuromuscular reflex activity in healthy adults. Hence, regardless of perturbation direction ventral muscles always revealed higher activity while compensating walking perturbations. The neuromuscular activation pattern reflects the intensity as well as the location of the perturbation: increased activity of the trunk muscles ipsilateral to the point of application of the perturbation. Therefore, the transfer of perturbations into sensorimotor training programs should consider a diversity of perturbations with respect to different neuromuscular response patterns of the trunk muscles.

ACKNOWLEDGEMENTS

* The present study was initiated and funded by the German Federal Institute of Sport Science and realized within MiSpEx – the National Research Network for Medicine in Spine Exercise.

(Granted number: BISp IIA1-080102A/11-14).

The present study was funded by the European Union (ERDF – European Regional Development Fund) (Granted number: 80132471).

CONFLICTS OF INTEREST

There is no conflict of interest.

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6. Are there differences in 3D trunk kinematics during continuous trunk loading while lifting objects with different weights?

Influence of load on 3-D segmental trunk kinematics in one-handed lifting: A Pilot Study

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Running Title: Trunk kinematics during lifting

Reference:

Müller J, Müller S, Stoll J, Rector M, Baur H, Mayer F.: Influence of load on 3-D segmental trunk kinematics in one-handed lifting: A Pliot Study. J Appl Biomech. 2016 Oct;32(5):520-5. doi: 10.1123/jab.2015-0227.

6.1 Abstract

Stability of the trunk is relevant in determining trunk response to different loading in everyday tasks initiated by the limbs. Descriptions of the trunk's mechanical movement patterns in response to different loads while lifting objects are still under debate. Hence, the aim of this study was to analyse the influence of weight on 3-D segmental motion of the trunk during one-handed lifting.

Ten asymptomatic subjects were included (29±3 yrs;1.79±0.09 m;75±14 kg). Subjects lifted 3x a light and heavy load from the ground up onto a table. 3-D segmental trunk motion was measured (12 markers; 3 segments: upper thoracic area [UTA], lower thoracic area [LTA], lumbar area [LA]). Outcomes were total motion amplitudes (ROM;[°]) for anterior flexion, lateral flexion and rotation of each segment.

The highest ROM was observed in the LTA segment (anterior flexion), and the smallest ROM in the UTA segment (lateral flexion). ROM differed for all planes between the three segments for both tasks ($p<0.001$). There were no differences in ROM between light and heavy load ($p>0.05$). No interaction effects (load*segment) were observed, as ROM did not reveal differences between loading tasks. Regardless of weight, the three segments did reflect differences, supporting the relevance of multi-segmental analysis.

Keywords: trunk motion, kinematic trunk model, everyday task, MiSpEx*

6.2 Introduction

One important task of the trunk during daily life activities is the compensation and stabilization of external loads initiated by the upper or lower limbs[1-5]. The analysis of trunk muscle activity is often used to describe stabilization strategies while compensating these external loads[6-9]. In addition, the analyses of trunk movement (kinematics) have been examined to gain additional insight into the movement patterns during loading, and are beneficial in emphasizing mechanical factors that may be associated with the development of back pain[4,10-13].

In the past, one rigid segment had been used to describe the motion of the trunk during different static and dynamic tasks[4,10,11,13]. These models, however, neglect the segmental differences between the thoracic and lumbar regions, which may be of interest when investigating the development of injuries and overload[14]. Nevertheless, recent studies have emphasized the need for a multi-segment trunk model in order to evaluate the characteristic trunk motion patterns during different, mostly static or not daily life tasks, mostly in healthy subjects[4,10,11,14]. These studies identified differences in the inter-segmental movements of the trunk, which is why Preuss et al. assessed the effect of segmentation on the total trunk motion measured in healthy adults during sitting and targeted leaning forward[11]. In their study, three kinematic models implementing either (1) one rigid segment, (2) three segments or (3) seven segments, were compared. The authors concluded that the kinematic model used would significantly influence the total trunk motion measured during multi-planar movements. Furthermore, they stated that a multi-segmental model allows differentiation between individual motion patterns. Nevertheless, the three-segment model described by Preuss & Popovich only partially covers the trunk, and therefore did not allow the interpretation of the movement of the lumbar, lumbar-thoracic and thoracic regions within the same model[11]. In addition, due to the nature of the sitting task analysed, the transferability to a more functional, dynamic daily life activity (e.g. lifting a box) must be questioned. Unfortunately, most of the previously mentioned kinematic trunk models lack information on their usage in more dynamic daily life activities[10,11].

In terms of lifting tasks, McGill et al.[4] investigated the influence of different weights (5 kg, 10 kg, 15 kg, 20 kg, 30 kg) and carrying conditions (one-handed vs. two-handed) on low back load. One-handed carrying led to greater low back loads compared to two-handed carrying of the same weight due to an increased shear stress on the spine. Additionally, they concluded that heavier weights resulted in greater low back loads. Therefore, one-handed lifting proposes a more challenging situation compared to two-handed lifting. Moreover, different loads might provoke different kinematic motion patterns of the trunk and its regions as part of the kinematic compensation strategy of the trunk, even in healthy subjects.

However, the influence of different loads on segmental trunk motion in healthy subjects during lifting remains unclear. Therefore, the purpose of this study was to compare three-dimensional segmental motions of the trunk during light and heavy loaded one-handed lifting in a healthy cohort, aiming to exclude the confounding variable of pain. It was hypothesized that motion amplitudes reduce from the thoracic to the lumbar region for both loading tasks. In addition, it is expected that light and heavy loading reflect different kinematic motion patterns, with heavy loading showing reduced motion amplitudes.

6.3 Material and methods

Subjects

Fifteen (7 male/ 8 female) adults with a minimum of 2 hours training per week were recruited out of a population of Master students at the University with the help of an informational flyer and personal contact. All participants read and signed an informed consent form before voluntarily participating in the study. The University of Potsdam´s Ethical Commission approved the study. Subjects reporting pain (visual analogue scale >0.5cm (VAS); [0-10cm][15]) previous to, during (N=3) or at the end (N=2) of the measurement protocol were excluded from the final analysis to avoid the possible influence of pain on the results. A total of 10 subjects (6 male / 4 female; 29 ± 3 yrs; 75 ± 14 kg; 1.79 ± 0.09 m; BMI 23.0 ± 2.3 kg/m²), free of complaints, general disease history and with no history of back pain, were included in the final data analysis. In accordance with Preuss & Popovich[11], this sample size was evaluated as sufficient to detect possible differences in kinematic trunk motion.

Measurement protocol

After assessing the anthropometrics, subjects were prepared for kinematic measurements. Twelve reflective markers were positioned over bony structures of the subjects' trunk (Fig. 9). Before the lifting tasks, every subject performed a 5-minute warm-up on a bicycle ergometer. Afterwards, subjects lifted, in random order, three times each, a light (1 kg) and heavy (10 kg;) load from the ground to the top of a table (height: 0.75 m) with the left hand. Subjects began all lifting tasks in an identical position, a hip-width bipedal upright stance. Subjects were instructed to lift the load with a self-selected moderate speed, starting with slight bending of the knees and the trunk (Fig. 10). Each lifting task was first demonstrated by the examiner, and subjects performed one test trial before starting the measurement.

Kinematic analysis

Segmental trunk motion was measured with a 3D-motion analysis system (8 cameras (MX3), 200Hz; Vicon Ltd., Oxford, UK). The kinematic trunk model used was adapted from Preuss & Popovich[11], and consisted of 12 markers framing three functional segments (upper thoracic area segment [UTA], lower thoracic area segment [LTA] and lumbar area segment [LA]). Reflective markers (diameter 9.5

mm) were positioned on the spinous processes of T3, T6, T9, T12, L3 and S1, and 0.05 m to the left and right of the spinous processes of C7, T6 and T12 (Fig. 2A) with double-sided tape. Markers in the middle of each triangle segment - T3 (UTA segment), T9 (LTA segment) and L3 (LA segment) - act as origin markers. The primary axis of each segment was formed between the origin marker and the perpendicular marker below (e.g. UTA segment: between T3 and T6). Finally, the plane was generated with the third marker in the upper right corner of each triangle segment (e.g. UTA segment: C7 right marker) which was connected to the segment. In accordance with Preuss & Popovich[11], a Cartesian coordinate system was created for each of the three trunk segments. The X-axis is running from posterior to anterior, the Y-axis in parallel to a line of the upper two marker from left to right and the Z-axis is running from caudal to rostral. All calculations of segment angles where done in relation to the global coordinate system (Cartesian system) following an orthopeadic convention using y as the first component followed by x and z.

To account for movement artefacts (e.g. extreme marker movement due to posture of the subjects spine or due to loss of halt on subjects skin), one investigator thoroughly observed the markers' motion throughout the lifting tasks during measurements, and problems were documented and considered for data analysis Calibration (5 Marker Wand & L-Frame, 200Hz; Vicon Nexus 1.8, UK) of the laboratory was conducted before each measurement. Marker data were analyzed (Vicon Nexus 1.8) to calculate the angles of each segment in relation to the laboratory coordinate system (global angle). The main outcome was total motion amplitude (ROM; [°]) assessed for anterior flexion, lateral flexion and axial rotation of each segment, a valid representative in the analysis of trunk motion in healthy and back pain patients[16,17]. The ROM was calculated between the time points of gripping the object from the ground (maximum bending) and placing it on the table (upright position). An exemplary kinematic motion curve is shown in Fig. 10. Positive values represent trunk anterior flexion, left-sided lateral flexion and left-sided rotation. In contrast, negative values characterize trunk extension, right-sided lateral flexion and right-sided rotation.

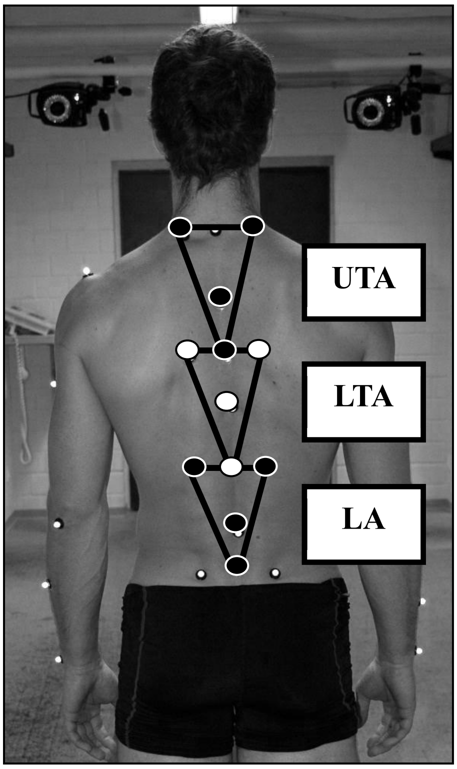
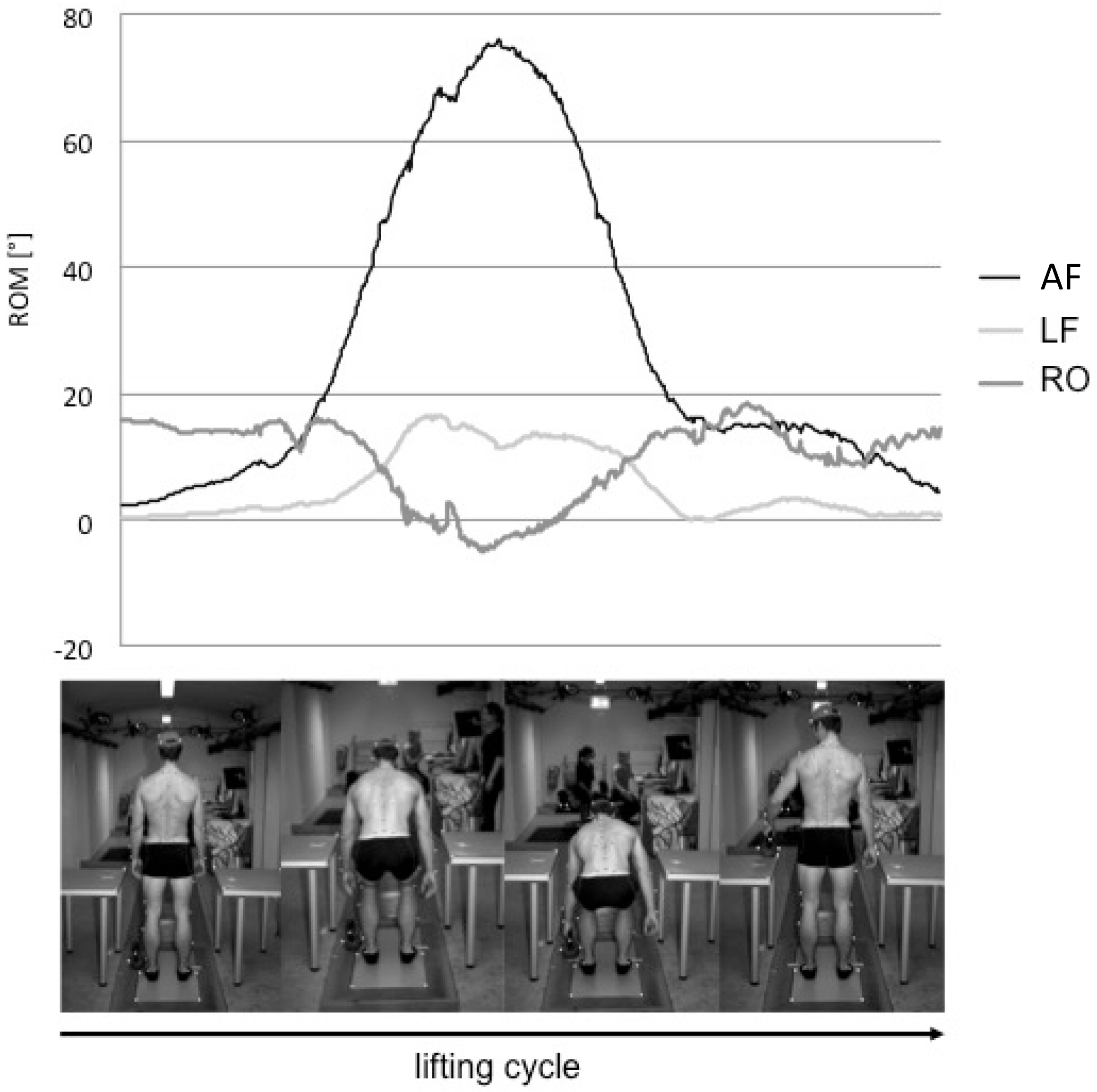


Figure 9: Kinematic trunk model (adapted/modified from Preuss & Popovich, 2009)

(segments: upper thoracic area [UTA], lower thoracic area [LTA], lumbar area [LA])



Legend:

Positive values: anterior flexion, left-sided lateral flexion and left-sided rotation.

Negative values: extension, right-sided lateral flexion and right-sided rotation.

Figure 10: Exemplary motion curve (ROM; N=1) of one lifting cycle for segmental trunk kinematics of LA segment in all planes (AF = Anterior Flexion; LF = Lateral flexion; RO = Rotation) while lifting a heavy load (10 Kg)

Data analysis:

All non-digital data were documented in a handwritten case report form (CRF) and transferred to a database (JMP Statistical Software Package 9, SAS Institute®). The main outcome measure – the total motion amplitude (ROM) for each segment and plane – was calculated as the mean ROM of the three repetitions for each lifting task.

Data were first evaluated descriptively, including means, standard deviations (SD) and 95%-confidence intervals (CI) for absolute ROM in each segment and motion plane for both lifting tasks (L; H) followed by two-way ANOVA (Factor: load (light; heavy)/ segment) analysis for dependent samples (post-hoc Tukey Kramer; $\alpha=0.05$) to compare segmental ROM differences between light and heavy loads. In addition a (post-hoc) power analysis was applied (JMP Statistical Software Package 9, SAS Institute®) to verify the results of the 2-way ANOVA including differences between segments and load*segment.

6.4 Results

The ROM did differ significantly between the three segments in all planes (anterior flexion; lateral flexion; rotation) in both lifting tasks (light; heavy) ($p<0.001$) (Fig. 11). In addition, comparing the ROM of all three segments (UTA; LTA; LA) in all planes between the light and heavy lifting tasks yielded no significant differences (anterior flexion: $p=0.7$; lateral flexion: $p=0.7$; rotation: $p=0.05$). No interaction effects (load*segment) were observed ($p=0.3$). Power (post-hoc) was 0.51 for the two-way-ANOVA (load*segment). In consideration of segment differences, power was 1.0.

The greatest ROM for both light and heavy lifting was observed in the LTA segment for flexion. The smallest ROM was found in the UTA segment for lateral flexion, again in both lifting tasks (Tab. 8).

Regardless of lifting task, the ROM in anterior flexion was significantly greater in the UTA and LTA segments compared to the LA segment ($p<0.001$). During lateral flexion and rotation, statistically significant differences were detected between the UTA segment and the two lower segments (LTA; LA)($p<0.001$), with the UTA segment having a significantly decreased ROM, comparatively. This finding was valid for both lifting tasks. (Fig. 11)

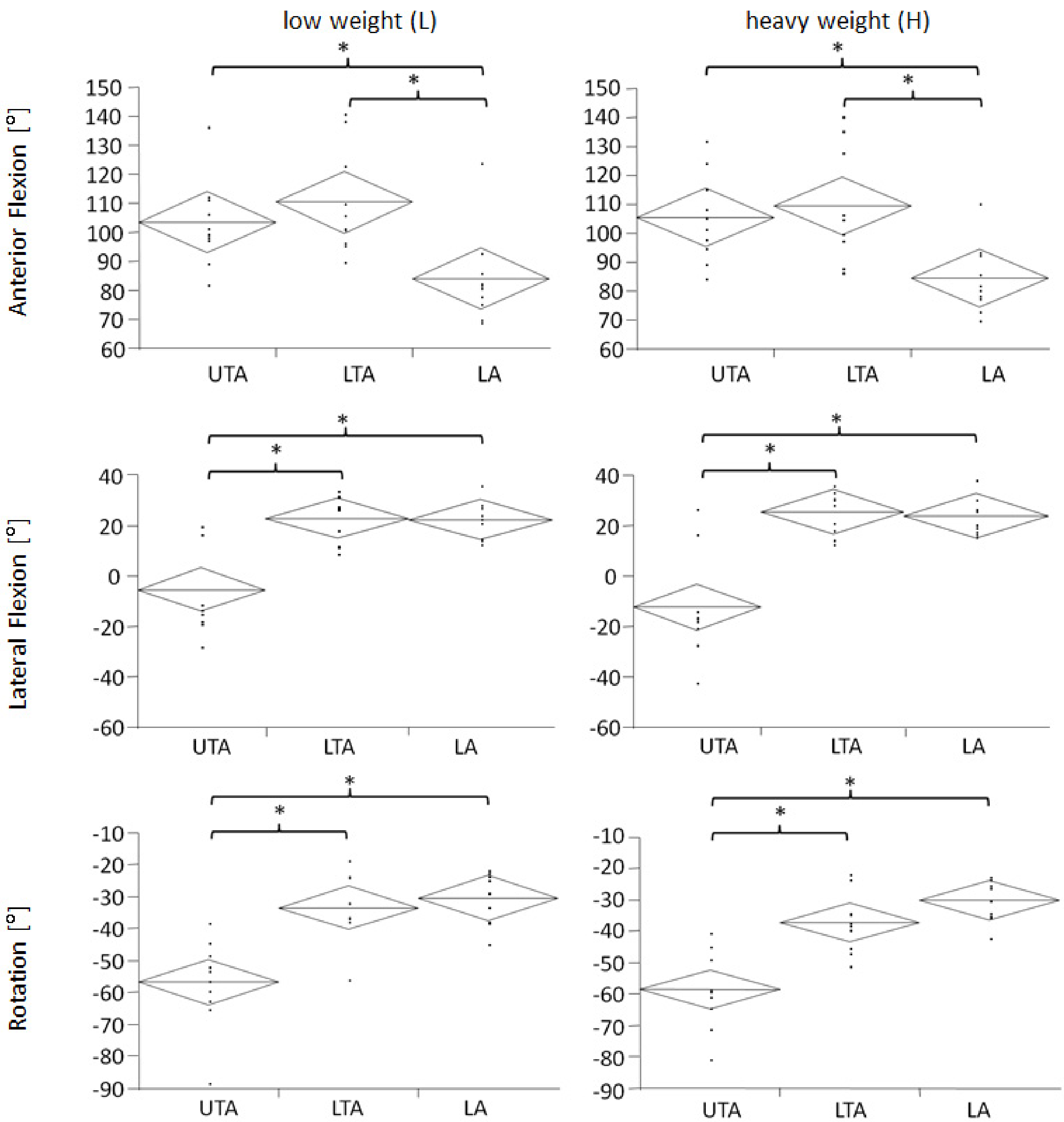


Figure 11: Absolute ROM with mean and 95% confidence interval for all segments in all planes for both lifting tasks (light and heavy) (* indicating a p<0.05)

(segments: upper thoracic area [UTA], lower thoracic area [LTA], lumbar area [LA])

Table 8: Absolute ROM (M1) reported for each segment in each plane for both lifting tasks (L and H) with mean ± standard deviation (± SD) [°] and p-values investigating differences between the lifting tasks.

	UTA					
	anterior flexion		lateral flexion		rotation	
	mean	SD	mean	SD	mean	SD
L	103.83	14.76	-5.01	18.69	-56.46	13.77
H	105.70	14.99	-12.05	21.24	-58.40	12.03
p-value	0.78	--	0.47	--	0.74	--
	LTA					
	anterior flexion		lateral flexion		rotation	
	mean	SD	mean	SD	mean	SD
L	110.51	18.01	23.33	9.48	-33.12	10.36
H	109.70	19.03	25.82	8.30	-36.81	9.49
p-value	0.92	--	0.54	--	0.42	--
	LA					
	anterior flexion		lateral flexion		rotation	
	mean	SD	mean	SD	mean	SD
L	84.33	15.73	22.57	7.13	-30.07	7.86
H	84.66	11.85	24.21	7.16	-29.73	6.47
p-value	0.96	--	0.61	--	0.92	--

Legend (segments): upper thoracic area [UTA], lower thoracic area [LTA], lumbar area [LA]

6.5 Discussion

The main purpose of this study was to investigate the influence of light and heavy loading on 3-dimensional segmental trunk motion during a one-handed lifting task. The main findings included (1) the elucidation of significant differences in segmental motion of the trunk independent of loading, but (2) an absence of trunk motion differences during light and heavy lifting tasks. The study was able to prove the relevance of a multi-segmental trunk analysis by use of a functional three-segmental model. In contrast, the differentiation between the loads was limited in a small healthy cohort.

Regarding the study's hypothesis, motion amplitudes were reduced only in the frontal plane from the thoracic to lumbar regions, but not for the sagittal and axial planes. Despite loading, the greatest motion amplitudes were found in anterior flexion, followed by moderate motion in left-sided rotation (positive values) in all segments. Lateral flexion showed the smallest amplitudes, with a left-sided deviation for the lower thoracic segment (LTA) and lumbar segment (LA) and a right-sided counter-motion (negative values) for the upper thoracic segment (UTA). In conclusion, particular movement patterns during lifting were observed between the three segments analyzed, independent of loading task and in line with literature using different segmentation[18,19].

The amplitude values presented in the study are in agreement with previously reported results[14]. In contrast, greater maximum amplitudes of the lumbar spine were observed compared to Bible et al.[20], which can be explained by the different objects used for the lifting task. Discrepancies in the maximal ROM between presented results and the literature could have been expected, considering the different data analysis (e.g. biomechanical model) and experimental tasks used. Nevertheless, the presented results reveal segmental differences between the three trunk areas regardless of loading task, which supports the need for multi-segmental trunk marker set-ups in the analysis of trunk motion, as suggested in recent literature analyzing fewer daily life tasks[4,11,14]. With respect to loading tasks, the expected differences in the kinematic motion patterns between light and heavy loading could not be shown for any segment or plane observed. Hence, according to our results it might be speculated that lifting, as an omnipresent task in daily life, might be interpreted as a stable movement pattern comparable to the human gait[4,21]. This could explain why no change in the kinematic lifting pattern with a heavier load (10 kg) was detected in the investigated cohort. The absence of kinematic differences between light and heavy loads in the presented cohort could be discussed as the result of an adapted neuromuscular activation pattern of the trunk's surrounding muscles[22,23]. Watanabe et al.[22] and Yoon et al.[23] reported increased muscular activity, especially in the back muscles, with greater loading during lifting tasks, assumingly to ensure trunk stability. Following this, it could be hypothesized that subjects stabilize their trunks with increased neuromuscular activity, aiming to keep a constant movement pattern[22]. An additional increase in loading may be necessary to overcome the individual limits of kinematic and neuromuscular trunk stability. One could speculate that these asymptomatic subjects were capable of compensating the additional 10kg load during this automated movement by means of active structures (trunk-surrounding muscles).

The limitations of this study should, however, be considered. The kinematic analysis was based on an examination of the ROM in relation to a global coordinate system. Therefore, a comparison of our results to other studies might be limited[11]. Nevertheless, the absolute ROM, based on global angle analysis, shows good to excellent reliability and is supported by the literature[14]. A global coordinate system enables a comparison between the movement patterns of 3 segments in 3 planes. Intersegmental angles (based on local coordinate systems) represent relative angles without reporting which segment had which amount/pattern of motion. Despite this, differences between segments of the trunk were obvious in the results of the study verified by statistical power analysis. In contrast, the small sample size has to be considered while interpreting the results of the analysis load*segment. During lifting, all subjects lifted the exact same weights (1 & 10 kg), irrespective of different anthropometrics (body height/mass). Furthermore, a standardized table height (0.75 m) was used regardless of subjects' body height. This is due to its comparability to certain daily life tasks (e.g. carrying a crate full of bottles). Therefore, no individual adaptations were made. Besides, the given instruction to the subjects on how to lift the objects might have influenced the results, leading to no differences between lifting tasks.

To conclude, the three trunk segments (UTA; LTA; LA) investigated support the relevance of a multi-segmental trunk analysis. The absence of differences in ROM between light and heavy loaded lifting trials may lead to the assumption that asymptomatic subjects are fully able to compensate (muscularly) an additional - expected and continuously applied - load without alterations in trunk motion. Higher loading – e.g. suddenly applied - seems to be necessary when investigating individual (kinematic/neuromuscular) limits of trunk stability. The effect of back pain on trunk motion patterns while lifting different loads and an influence on therapy and rehabilitation content needs to be investigated in future research.

ACKNOWLEDGEMENTS

The authors thank G. Hain and A. Reschke for helping with data assessment and analysis. The authors appreciated the assistance of M. Löhrer (prophysics, Switzerland) with the development of the kinematic trunk model.

*The present study was initiated and funded by the German Federal Institute of Sport Science and realized within MiSpEx – the National Research Network for Medicine in Spine Exercise.

(grant number: BISp IIA1-080102A/11-14).

The present study was also funded by the European Union (ERDF – European Regional Development Fund; grant number: 80132471).

CONFLICTS OF INTEREST

There is no conflict of interest.

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7. Does sudden loading (perturbation) while walking affect 3D trunk kinematics and neuromuscular control in healthy individuals?

Stumbling reactions during perturbed walking: Neuromuscular Reflex Activity and 3-D Kinematics of the Trunk – A Pilot Study

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Article type: Original Investigation

KEYWORDS: trunk kinematics, treadmill walking, gait perturbation, EMG

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Müller J, Müller S, Engel T, Reschke A, Baur H, Mayer F. Stumbling reactions during perturbed walking: Neuromuscular reflex activity and3-D kinematics of thetrunk – A pilot study. J Biomech. 2016;49(6):933-8. doi: 10.1016/j.jbiomech.2015.09.041.

7.1 Abstract

Reflex activity of the lower leg muscles involved when compensating for falls has already been thoroughly investigated. However, the trunk's role in this compensation strategy remains unclear. The purpose of this study, therefore, was to analyze the kinematics and muscle activity of the trunk during perturbed walking. Ten subjects (29±3yrs;79±11cm;74±14kg) walked (1m/s) on a split-belt treadmill, while 5 randomly timed, right-sided perturbations (treadmill belt deceleration: 40m/s^2) were applied. Trunk muscle activity was assessed with a 12-lead-EMG. Trunk kinematics were measured with a 3D-motion analysis system (12 markers framing 3 segments: upper thoracic area (UTA), lower thoracic area (LTA), lumbar area (LA)). The EMG-RMS [%] (0-200ms after perturbation) was analyzed and then normalized to the RMS of normal walking. The total range of motion (ROM;[°]) for the extension/flexion, lateral flexion and rotation of each segment were calculated. Individual kinematic differences between walking and stumbling [%; ROM] were also computed. Data analysis was conducted descriptively, followed by one- and two-way ANOVAs (α=0.05).

Stumbling led to an increase in ROM, compared to unperturbed gait, in all segments and planes. These increases ranged between 107±26% (UTA/rotation) to 262±132% (UTS/lateral flexion), significant only in lateral flexion. EMG activity of the trunk was increased during stumbling (abdominal: 665±283%; back: 501±215%), without significant differences between muscles.

Provoked stumbling leads to a measurable effect on the trunk, quantifiable by an increase in ROM and EMG activity, compared to normal walking. Greater abdominal muscle activity and ROM of lateral flexion may indicate a specific compensation pattern occurring during stumbling.

KEYWORDS: trunk kinematics, treadmill walking, gait perturbation, EMG

7.2 Introduction

Compensation during unexpected full-body perturbations is an essential mechanism for achieving the postural control needed in daily life activities as well as high-performance sports (Hodges et al. 2001; Kibler et al. 2006). One example of an unexpected, high-loading situation is the compensation to prevent falls after slipping or tripping, particularly common in elderly populations (Cholewicki et al. 2000; Cordero et al. 2003; Ferber et al. 2002; Grabiner et al. 1993; Granacher et al. 2010a). Even in elite athletes, the compensation for unexpected lower leg perturbations (e.g. cutting/stopping maneuvers) is essential for achieving stability, continual performance (Kibler et al. 2006) and injury risk reduction (Hibbs et al. 2008). Previous studies have shown that reflex activity of the lower leg muscles is involved in the compensation of lower leg perturbations (Dietz et al. 1989; Dietz et al. 2004; Keck et al. 1998; Ferber et al. 2002; Oliveira et al. 2012; Tang et al. 1998; Taube et al. 2007a).
However, it remains unclear whether a perturbation of the lower extremities leads to a displacement of the trunk, one possible relevant factor in the compensation strategy. One main function of the trunk is the compensation of external forces in order to ensure stability of the body (Cresswell et al. 1994; Kibler et al. 2006). In order to investigate these compensation strategies for external perturbations, the activity and coordination of the trunk’s surrounding muscles were analyzed (Cresswell et al. 1994; McGill et al. 2013; Radebold et al. 2000; Stanek et al. 2011; Tang et al. 1998; Thomas et al. 1998). Most of these studies applied loads directly to the trunk, with one exception in which loads were applied to the upper and lower limbs (Hodges et al. 2001) in less dynamic situations (e.g. standing or sitting) (Cresswell et al. 1994; Moseley et al. 2002; Radebold et al. 2000; Vera-Garcia et al. 2007). Therefore, transferability of these results to dynamic situations could be discussed critically. To the authors’ knowledge, no study has yet investigated the kinematics and muscular activity of the trunk during high-intensity perturbation while walking, a more functional situation. Hence, the purpose of this study was to investigate whether muscular activity and three-dimensional kinematics of the trunk contribute to the compensation strategy used during perturbed treadmill walking. The authors expected relevant alterations during stumbling compared to normal walking in the activity of the trunk’s surrounding muscles, as well as a kinematic trunk response to the perturbation. Moreover, identification of typical muscular and kinematics responses was expected, which could assist in the development of fall-prevention interventions or core stability programs.

7.3 Material and methods

Subjects

Ten adults, free of complaints, no history of back pain and with a minimum of 2 hrs of physical activity per week were included in the study (5 male / 5 female; 29 ± 3 yrs; 179 ± 11 cm; 74 ± 14 kg). All participated voluntarily after giving written informed consent. The University of Potsdam's ethics committee approved the study.

Measurement protocol

Subjects were prepared for EMG and kinematics before a short warm-up and familiarization: 5 minutes of walking without perturbation (1m/s) on a split-belt treadmill (Woodway, Weil am Rhein, D; Fig. 12/13A). Afterwards, subjects continued to walk at a baseline velocity of 1 m/s, a typical walking speed in adults (Winter & Yack 1987), while five randomly timed, right- and left-sided perturbations were applied 200 ms after initial heel contact, triggered by a plantar pressure insole (Pedar X, Novel, Munich, D). During perturbations, one belt decelerated (amplitude = 2 m/s; deceleration = 40 m/s^2) for 50 ms, before returning to baseline after an additional 50 ms (Fig. 13B). Cordero et al. (2003) explained that more than one stride was necessary for recovery after gait perturbation. Therefore, a minimum of 10 seconds of rest between perturbations ensured that participants had resumed their normal walking patterns. For safety reasons, participants were provided with a waist belt connected to an emergency stop release. Kinematics and EMG of the trunk were assessed during normal (unperturbed) gait (G) and stumbling (S). Only right-sided perturbations were used in the final data analysis due to direct triggering of the perturbations by the plantar pressure insole. Left-sided perturbations were applied to ensure that participants did not adapt their normal walking pattern to only right-sided perturbations.

Electromyographic analysis

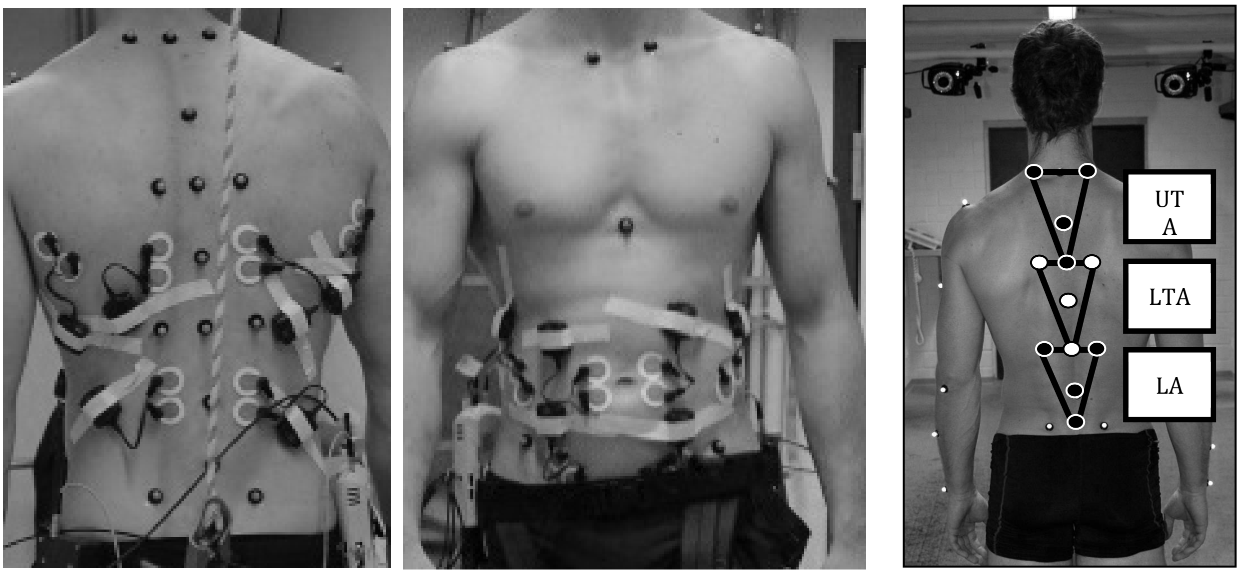
Muscular activity of the trunk was assessed using a 12-lead surface EMG (Radebold et al. 2000) (Fig. 12A): Mm. rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right sides; Mm. erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of left and right sides. Muscular activity was analyzed using bilateral and bipolar surface telemetry EMG (band-pass filter: 5 Hz to 500 Hz, gain: 5.0, overall gain: 2500, sampling frequency: 4000 Hz, RFTD-32, myon AG, Baar, CH). The localization of electrodes was carefully determined according to Radebold et al. (Radebold et al. 2000). Before electrodes (AMBU Medicotest, Denmark, Type N-00-S, interelectrode distance: 2 cm) were applied, the skin was shaved and slightly roughened to remove surface epithelial layers and control skin resistance (<5 kΩ). The longitudinal axes of the electrodes were placed in line with the underlying muscle fibers. The signal was rectified and averaged before calculation of the amplitudes. The first 5 unperturbed strides and the 5 perturbed strides were averaged to one mean unperturbed and perturbed step, followed by the root mean square analysis of the amplitudes. Root mean square values (RMS; [%]) served as the main outcome measure and were analyzed within the subsequent 200 ms after perturbation to ensure the analysis of reflex activity of the trunk's surrounding muscles (Granacher et al. 2010a; Taube et al. 2007). RMS was normalized to RMS of the same interval during unperturbed walking (Granacher et al. 2010a; Taube et al. 2007) (Fig. 14). Additionally, the mean (normalized) EMG-RMS of four trunk areas was calculated: right abdominal area (AR; Mm RA, EO, IO of right side); left abdominal area (AL; Mm RA, EO, IO of left side); right back area (BR; Mm UES, LES, LD of right side) and left back area (BL; Mm UES, LES, LD of left side) (McGill et al. 2013).

Kinematic analysis

Segmental trunk motion was measured with a 3D-motion analysis system (8 cameras (Vicon MX); 200Hz; Vicon Ltd., Oxford, UK) and the kinematic trunk model from Preuss et al. (Preuss and Popovic 2010) was adapted with only slight modifications. In detail, 12 markers were used to frame three segments (upper thoracic area (UTA), lower thoracic area (LTA) and lumbar area (LA)) (Fig. 12B). Reflective markers (diameter: 9.5 mm) were positioned on the spinous processes of T3, T6, T9, T12, L3, and S1 and 0.05 m left and right of the spinous processes of C7, T6 and T12 with double-sided tape. Marker data were analyzed (Vicon Nexus 1.8) to calculate the angles of each segment in relation to the laboratory coordinate system (global angle). Helical axes, without rotation sequences, were used. The main outcomes were the total motion amplitudes (ROM; [°]) for the extension/flexion (E/F), lateral flexion (LF) and axial rotation (RO) of each segment. The mean ROM during normal unperturbed gait (G) was calculated from the first five unperturbed steps of the protocol. The mean ROM during perturbed walking (S) was calculated from the 5 repetitions of right-sided perturbations. Additionally, individual kinematic differences between walking and perturbed walking [S in % of G; ROM] were computed for the extension/flexion (E/F), lateral flexion (LF) and axial rotation (RO) of each segment.

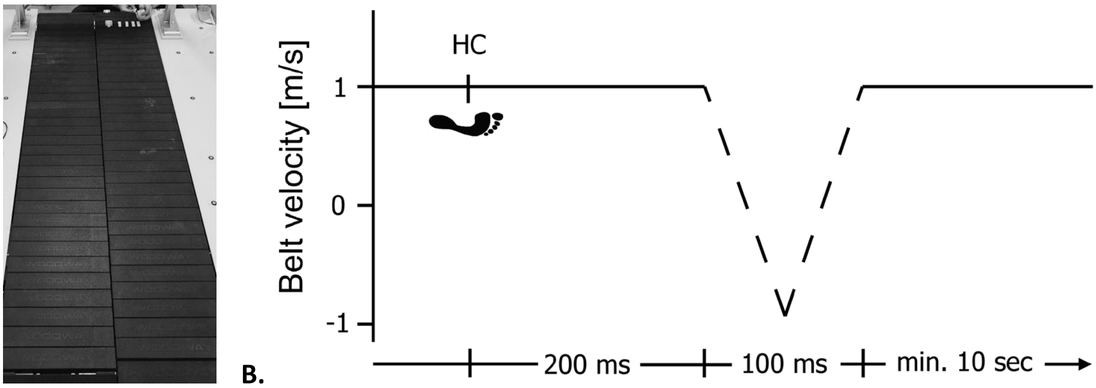
Data analysis:

Data were descriptively analyzed, including means and standard deviations (SD), of EMG (RMS %) and kinematics (absolute ROM in each segment and motion plane) for normal gait (G) and perturbed walking (S). In addition, a one-way ANOVA was applied to test for EMG-RMS differences (dependent variable) between all muscles and between the four trunk areas (independent variables). Kinematics (dependent variable) were tested with a one-way ANOVA for comparison between conditions (independent variable: normal gait/stumbling). For analysis of interaction effect between segments and planes (independent variables) a two-way ANOVA was applied (dependent variable: ratio of angles between S and G). The level of significance was set to α=0.05.



A B

Figure 12: A. 12-channel EMG-trunk-setup (Radebold 2002), B. Kinematic trunk model (adapted after Preuss et al. (2009)) with 3 segments: upper thoracic area (UTA), lower thoracic area (LTA), lumbar area (LA)



A. B.

Figure 13: A. Customized Split belt treadmill with 2 separate selectable belts (Woodway); B. Treadmill perturbation characteristics (HC: initial heel contact)

7.4 Results

The analysis of trunk muscle activation showed an increased RMS during stumbling (S) compared to normal walking (G) for both abdominal and back muscles, without a statistically significant difference between muscles (p=0.12) (Fig. 15A). On average, abdominal muscle RMS increased 665 ± 283 %, with RA of the right side showing the highest increase (902 ± 877 %). Back muscle EMG-RMS increased by 501 ± 215 % during S compared to G, with the largest increase of 703 ± 239 % found in the LES of the right side. The analysis of the trunk showed greater abdominal muscle activity during stumbling compared to the back muscles, with a greater increase in activity of the right abdomen (Fig. 15B). However, no statistically significant differences could be found between the four trunk areas (p=0.08) and between the abdominal and back muscles (p=0.053). The muscles on the right side (646 ± 348 %) showed a slightly greater increase in RMS compared to those of the left side (497 ± 248 %) without statistical significance (p=0.13).

Absolute values for the absolute motion amplitude (ROM) for each segment in each plane are shown in Table 9. Fig. 14B shows by way of example the range of motion over the whole stride cycle in comparison to unperturbed and perturbed strides for LF in LA segment. All 3 segments revealed greater absolute lateral flexion ROM during stumbling compared to normal walking, with statistical significance (LA: p=0.0001; LTA: p=0.0002; UTA: p=0.0016). No significant differences could be found for sagittal and transverse planes (p-values between 0.17 and 0.60).

Differences between S and G (Table 9) ranged between 107 ± 26 % (UTA in R) to 262 ± 132 % (UTA in LF). Further analysis revealed no interaction effect between the 3 segments and motion planes (p > 0.05). The greatest differences (approx. 2.5 times greater) were found in the lateral flexion of all segments. Axial rotation and extension/flexion revealed only small differences of less than 2.2 ° for all segments.

Table 9: Results for absolute ROM and EMG-RMS for normal walking (G) and stumbling (S) with mean ± standard deviation (± SD) and differences [%] between S/G

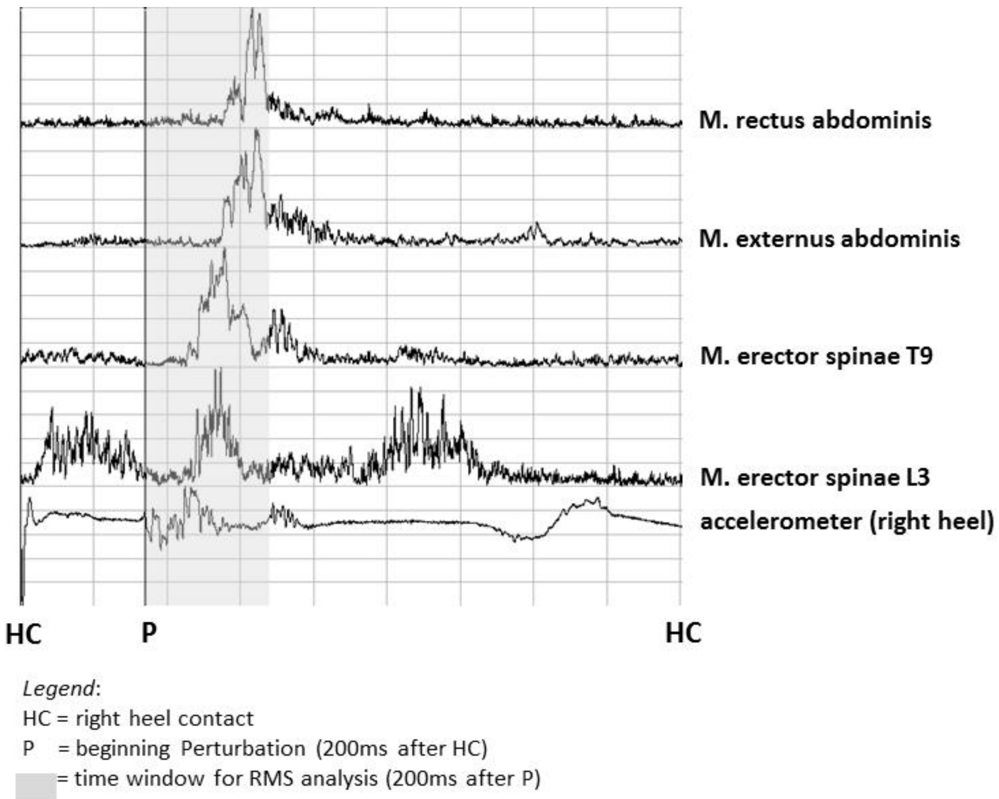
A Absolute ROM for each segment in each motion plane [°]

segment	plane	G [°]		S [°]		difference [S in % of G]
		mean	SD	mean	SD	S/G (mean±SD)
UTA	E/F	5.8	2.6	7.4	2.5	135 ± 47
	LF	3.8	0.9	9.4	4.7	262 ± 132
	RO	12.8	2.9	13.7	4.7	107 ± 26
LTA	E/F	6.9	1.5	8.3	2.7	121 ± 37
	LF	3.8	1.1	9.2	3.6	248 ± 67
	RO	12.6	3.3	14.3	4.2	115 ± 25
LA	E/F	6.0	1.2	7.0	2.8	123 ± 51
	LF	3.4	1.1	7.7	2.2	238 ± 78
	RO	13.9	3.4	16.1	4.6	117 ± 24

Legend: segment: UTA: upper thoracic area, LTA: lower thoracic area, LA: lumbar area
plane: E/F: extension/flexion, LF: lateral flexion, RO: axial rotation

B Absolute EMG-RMS amplitudes for each muscle [V]

muscles	G [V]		S [V]		difference [S in % of G]
	mean	SD	mean	SD	S/G (mean±SD)
M. rec. abd. ri	0.00998	0.00206	0.08436	0.07014	902 ± 872
M. rec. abd. le	0.00959	0.00261	0.05884	0.03388	670 ± 482
M. obl. ext. abd. ri	0.01907	0.00799	0.15540	0.09476	815 ± 520
M. obl. ext. abd. le	0.02924	0.01644	0.18288	0.08620	699 ± 317
M. int. abd. ri	0.02051	0.01004	0.08252	0.04435	399 ± 114
M. int. abd. le	0.01898	0.00731	0.09021	0.04952	506 ± 295
M. latis. dorsi ri	0.01274	0.00352	0.07402	0.04047	554 ± 245
M. latis. dorsi le	0.01823	0.00942	0.06994	0.03413	411 ± 247
M. erec. spinae TH9 ri	0.02016	0.00817	0.08882	0.03214	504 ± 285
M. erec. spinae TH9 le	0.02174	0.00954	0.08644	0.04816	471 ± 350
M. erec. spinae L3 ri	0.01645	0.00958	0.09020	0.05999	703 ± 658
M. erec. spinae L3 le	0.02420	0.01697	0.06790	0.03487	365 ± 265



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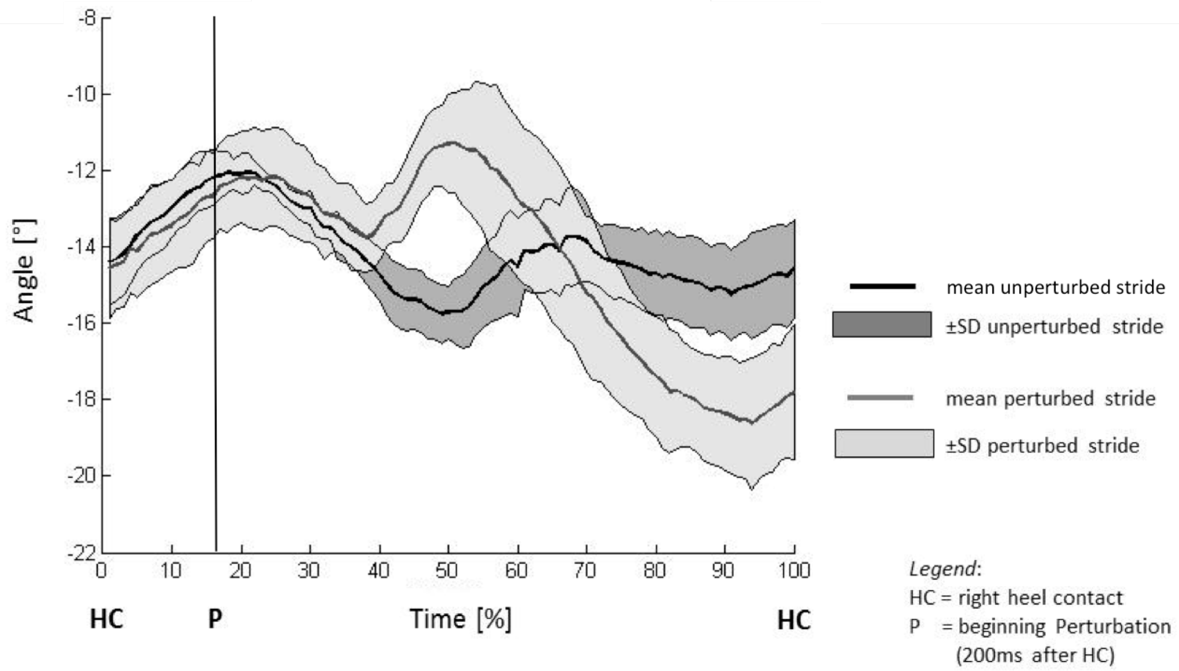
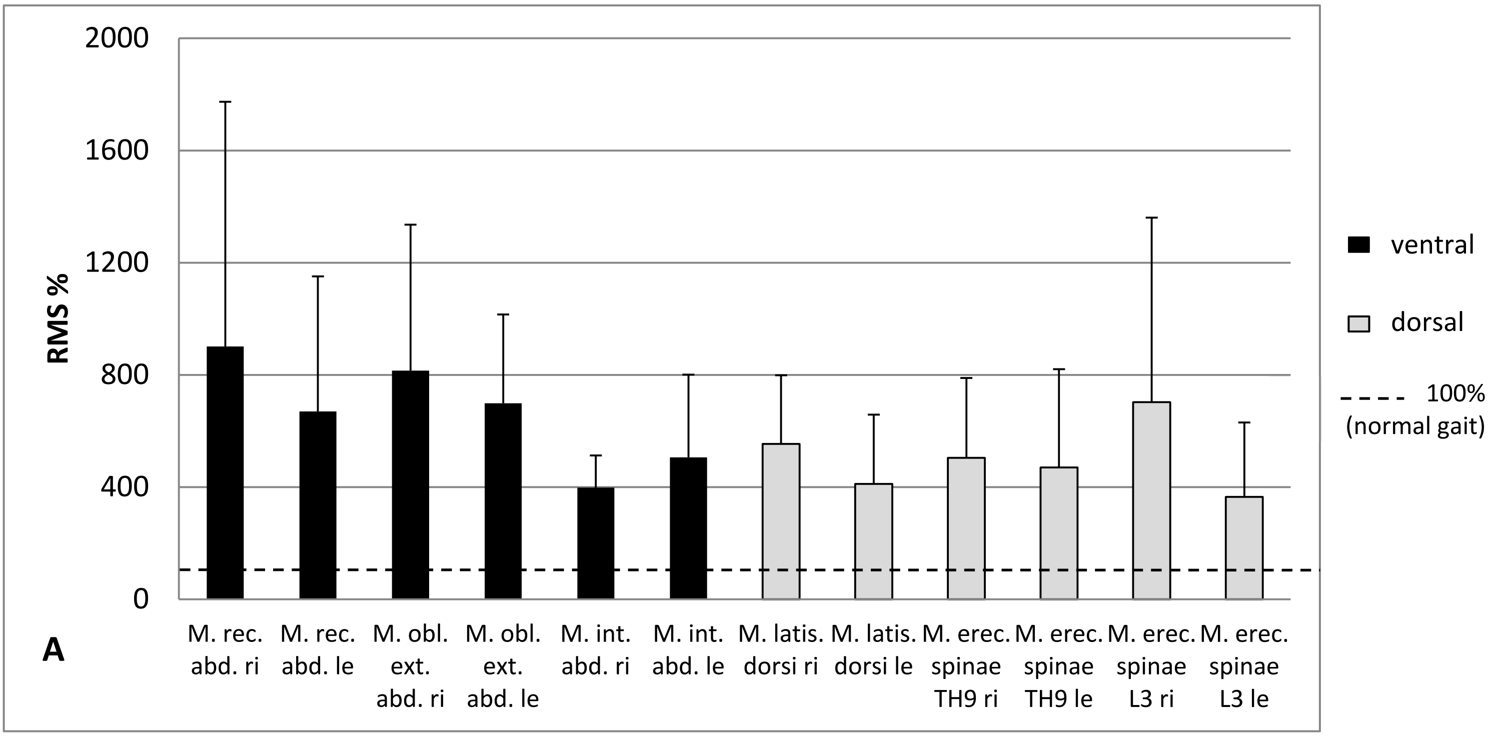


B

Figure 14: EMG and kinematic pattern of the trunk during perturbation

A EMG pattern of 4 trunk muscles (averaged, rectified signal of 5 perturbed steps for 1 subject)

B Kinematic pattern of the LA segment in lateral flexion: Comparison of unperturbed and perturbed step (mean ± SD for N=10 subjects)



B

AR: abdominal right (RA, EO, IO of right side); AL: abdominal left (RA, EO, IO of left side); BR: back right (UES, LES, LD of right side); BL: back left (UES, LES, LD of left side)

Figure 15: EMG-RMS for the trunk muscles (normalized to unperturbed walking): A. EMG-RMS [%] for six abdominal and six back muscles and B. EMG-RMS [%] for 4 areas of the trunk according to McGill (2013)

7.5 Discussion

Provoked stumbling led to an increase in trunk muscle activity, with no significant differences between back and abdominal or right- and left-sided trunk muscles. Furthermore, only changes in (left-sided) lateral flexion were found to be significant. Nevertheless, a small increase in left-sided axial rotation combined with a small increase in flexion of all segments could be shown.

Electromyographic activity of the trunk

The increased muscular activity of the trunk during perturbed gait in comparison to normal walking agrees with studies from Stanek et al. (Stanek et al. 2011) and Sakai et al. (Sakai et al. 2008), which found up to 4 times greater EMG activity (average EMG; peak EMG) of the Mm. obl. ext. abd. during perturbed gait compared to normal walking. In our study, EMG activity of the Mm. obl. ext. abd was nearly 7 times greater during stumbling. The comparison of abdominal and back muscles revealed greater EMG activity in the abdomen, in accordance with Cresswell et al. (Cresswell et al. 1994). Although no significant differences were found, a 150 % greater increase in abdominal muscles during stumbling must be considered relevant and thus part of a specific compensation pattern. On the one side, greater increases in abdominal muscle activity could be due to the application of a backwards perturbation, used to imitate slipping events in the past (Dietz et al. 1987; Grabiner et al. 1993; Granacher et al. 2010b). These new results then emphasize the importance of the perturbation direction (forward/backward). One could speculate that more back muscle activity would not actually be beneficial and would instead decrease the stability of the body due to trunk extension and the elevation of its center of mass (Grabiner et al. 1993). In contrast, it has been discussed that greater back muscle activity increases spine stiffness, thus enhancing trunk stability. Furthermore, the right side (both back and abdominal muscles) always exhibited greater increases (> 150 %) in contrast to the left side, albeit without statistical significance. Consequently, the muscles of the perturbed side may play a more substantial role when compensating for the subsequent loss of balance (McGill et al. 2013).

Kinematics of the trunk

A quantifiable displacement of the trunk during stumbling is in line with previous studies (Grabiner et al. 1993; Tang et al. 1998). It could be speculated that subjects aimed to regain body stability using specific compensation strategies, mainly by bending the trunk in the frontal plane (lateral flexion) (Grabiner et al. 1993; Tang et al. 1998). On the contrary, motion in the sagittal and transverse planes remained relatively constant. According to these data, lateral flexion could be discussed as the key kinematic element of trunk compensation during provoked treadmill stumbling. This is in contrast to Grabiner et al. (1993), who observed a dominance of trunk motion in the sagittal plane during perturbation in the anterior direction. They found 4-fold increases in trunk flexion compared to normal walking and discussed this as a decrease in postural stability, increasing the risk of falling (Grabiner et al. 1993). Differences in the methodological characteristics of the perturbation (slipping vs. getting caught by one foot) may have led to the different results. Since Grabiner et al. (Grabiner et al. 1993) argues that increased trunk flexion is adverse to achieving stability and lowering the center of mass, the recent cohort presented an appropriate response to the perturbation in order not to fall.

Limitations

During stumbling, all subjects walked at the same baseline velocity, despite different anthropometrics (body height, body mass) and stride parameters (individual preferred walking velocity; individual stride length). Moreover, the influence of walking speed on plantar pressure (Taylor et al. 2004), neuromuscular activity (Chumanov et al. 2007) and lower limb kinematics has been thoroughly investigated (Kang 2007). As a result, for a standardized and comparable test situation between all subjects, a consistent velocity during the stumbling protocol was favored (Cordero et al. 2003). Nevertheless, it cannot be ruled out that at this consistent velocity subjects were stressed to different extents.

The kinematic analysis was based on an examination of the range of motion in relation to a global coordinate system. A global coordinate system enables the comparison between movement patterns of all 3 segments in all 3 planes. Intersegmental angles (based on local coordinate systems) represent relative angles without reporting which segment had which degree/pattern of motion.

The EMG RMS results might be affected by the latencies of the trunk muscles. If latency is different between muscles, for example, a longer reflex latency leads to a shorter activity phase of the muscle analyzed during the fixed RMS time window (200 ms). The latencies of the trunk muscles in our study ranged between 82 ± 7 ms (ES T9) and 137 ± 25 ms (RA.).

In conclusion, provoked stumbling leads to a displacement of the trunk, kinematically measured in all 3 planes, and to an increase in trunk muscle EMG activity. A significantly greater lateral flexion ROM may indicate a specific compensation pattern in response to stumbling in healthy adults. Consequently, the derivation of complex fall prevention and/or core stability interventions should be comprised of exercises which address the lateral part of both abdominal and back muscles. Moreover, perturbations should be implemented into the exercises with focus on trunk muscle ipsilateral to the point of application of the perturbation.

ACKNOWLEDGEMENTS

The present study was initiated and funded by the German Federal Institute of Sport Science and realized within MiSpEx – the National Research Network for Medicine in Spine Exercise.

(Granted number: BISp IIA1-080102A/11-14).

The present study was funded by the European Union (ERDF – European Regional Development Fund).

CONFLICTS OF INTEREST

There is no conflict of interest.

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8. How does sudden loading affect trunk stability (quantified by 3D motion and neuromuscular activity) in healthy individuals and back pain patients?

Effects of sudden walking perturbations on neuromuscular reflex activity and three-dimensional motion of the trunk in healthy controls and back pain patients

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Müller J, Engel T, Müller S, Stoll J, Baur H, Mayer F. Effects of sudden walking perturbations on neuromuscular reflex activity and three-dimensional motion of the trunk in healthy controls and back pain patients. (under review Plos One 2016).

8.1 Abstract

It has been shown that the neuromuscular activity patterns of trunk muscles in back pain (BP) patients show an altered neuromuscular activity when artificial quasi-static sudden trunk loading is compared to healthy controls (H). Back pain patients (BPP) show a delayed muscle onset, increased co-contractions, and increased EMG variability. However, it is unclear whether these results can validly be transferred to suddenly applied walking perturbations, an automated but more functional and complex movement pattern. Moreover, there is an evident need to develop research-based strategies for the rehabilitation of back pain. Therefore, the investigation of differences in trunk stability between H and BPP in functional movements is of primary interest in order to define suitable intervention regimes. Therefore, the purpose of this study was to analyse whether neuromuscular reflex activity as well as three-dimensional trunk kinematics differ between H and BPP during sudden loading while walking.

Eighty H (31m / 49f; 29±9yrs; 174±10cm; 71±13kg) and 14 BPP (6m / 8f; 30±8yrs; 171±10cm; 67±14kg) walked (1m/s) on a split-belt treadmill while 15 right-sided perturbations (belt decelerating, 40m/s^2, 50ms duration; 200ms after initial heel contact) were randomly applied. Trunk muscle activity was assessed using a 12-lead EMG. Trunk kinematics were measured using a 3-segment model consisting of 12 markers (upper thoracic area (UTA), lower thoracic area (LTA), lumbar area (LA)). EMG-RMS ([%], 0-200ms after perturbation) was calculated and normalized to the RMS of unperturbed gait. In addition, latency (T_{ON}; ms) and time to maximum activity (T_{MAX}; ms) were analysed. Total range of motion (ROM;[°]) and mean angle (A_{mean};[°]) for anterior flexion, lateral flexion and rotation were calculated (time interval: whole stride cycle and 0-200ms after perturbation) for each of the three kinematic segments during unperturbed and perturbed gait. Data was analysed descriptively (mean±SD) followed by a student´s t-test to account for differences between H and BPP. The coefficient of variation (CV; [%]) was calculated for EMG-RMS and ROM to evaluate variability between the 15 perturbations for all groups (H/$H_{matched}$ vs. BPP). With respect to the unequal distribution of subjects to H and BPP, an additional matched-group analysis was conducted and 14 healthy controls were gender-, age- and anthropometrically matched (group $H_{matched}$) to the BPP.

No statistically significant group differences were observed for EMG-RMS, ROM, or CV (EMG/ROM) (p>0.025). BPP showed an increased T_{ON} and T_{MAX} compared to H ($H_{mtached}$), being statistically significant for Mm. rectus abdominus (p=0.019) and erector spinae T9/L3 (p=0.005/p=0.015). Range of motion analysis over the whole stride cycle and reflex response revealed no significant differences between groups (p>0.025). Regarding overall posture (A_{mean}), BPP showed a significant higher flexed posture of the UTA and LTA segment during normal walking. Trunk posture analysis (A_{mean}) during reflex response showed higher trunk extension (E/F) values in LA and LTA segments for H/$H_{matched}$

compared to BPP. In addition, matched group (BPP vs. $H_{matched}$) analysis did not show any systematic changes of the results (EMG/kinematics) between groups.

It can be summarized that back pain patients present a different neuromuscular and kinematic compensation strategy compared to healthy controls when unexpected sudden loading is applied during gait. In BPP muscles response times and overall trunk posture, especially in the sagittal and transversal planes, are impaired. This pattern could indicate a reduced trunk stability and higher loading during gait perturbations.

KEYWORDS: trunk, core, gait perturbation, kinematics, EMG, MISPEX*

8.2 Introduction

Non-specific back pain (BP) is a major burden on health systems of western societies, with a lifetime prevalence of about 85% and frequently leading to disability in 10% to 15% of all patients affected [1-4]. In etiology, potential causes for back pain are discussed including repetitive micro trauma and insufficiency of the muscle-tendon complex based on inadequate postural and neuromuscular control, reduced maximum trunk strength capacity and trunk muscle fatigue during dynamic loading [5-7]. In addition, these factors are defined important contributing to the stability of the trunk [8-11]. Therefore, great emphasis has been placed on the importance of trunk stability, especially in situations requiring compensation of (unexpected) high loading induced e.g. by perturbations [8-11]. Stability provided by the trunk muscles is considered meaningful in counteracting sudden, unexpected loading during daily life as well as dynamic, high-intensity activities [8,12]. Hence, optimizing neuromuscular core stability is considered beneficial for protection against sudden, repetitive and excessive overloading of the trunk [8,9,12,13,14,15].

When compensating for sudden external (un-)expected perturbations, delayed muscle onset, increased co-contractions, and increased EMG variability could be shown in back pain patients (BPP) [9,16-18]. However, most of the studies applied the load directly to the trunk, mainly in non-dynamic situations (e.g. standing or sitting) [17,19]. Therefore, the transferability of these results to dynamic loading, daily life or sports situations applied by the lower limbs cannot be validated, and has to be discussed critically due to the quasi-static and limited functional load application. Sudden loading during gait, therefore, might be a more suitable situation in which to analyse differences between healthy controls (H) and back pain patients (BPP) [20-23]. The human gait is described as an automated and stable movement pattern (high intra-individual reproducibility) with more functional and complex demands on the neuromuscular system and kinematic chain of the trunk compared to the quick-release experiments. Moreover, there is an evident need to develop research-based strategies for the prevention and rehabilitation of back pain. Therefore, the investigation of differences in trunk function and stability between healthy and back pain patients in functional movements is of primary interest in order to define adequate intervention regimes.

The analysis of trunk kinematics and posture comparing patients and healthy subjects has been discussed as beneficial for extracting the mechanical factors that may be associated with the development, persistence and recurrence of back pain [13,14,20]. However, inconsistent results regarding movement patterns and kinematic variability during gait have been found [24-27]. Vogt et al. (2001) [27] reported a higher stride-to-stride variability of all lumbar movement planes in lower back pain patients, while the absolute range of motion was unchanged compared to healthy controls. In addition, Steel et al. (2014) [26] reported a higher movement variability during gait in patients compared to healthy controls only in the sagittal and transverse planes. Moreover, some studies reported that symptomatic subjects display reduced lumbar rotational movements [20], while others showed that lower BP increases spine or pelvis rotation [28].

In summary, it is unclear whether back pain patients (BPP) suffer from delayed muscle reflex response and higher trunk movement variability with sudden dynamic loading during gait. Consequently, the purpose of this study was to analyse the effects of sudden walking perturbations on neuromuscular reflex activity and three-dimensional motion of the trunk in healthy controls and back pain patients. It is hypothized that BPP have increased reflex response times to sudden loading while walking with increase neuromuscular activity. In addition, an increased range of motion with higher movement variability in BPP is expected.

8.3 Material and methods

Subjects

The investigation was conducted at the University Outpatient Clinic and participants were recruited from the Outpatient Clinic (e.g. students and/or academic workers from the university population receiving physical examination, recreational athletes receiving annual health check-ups), using flyers (displayed at the university cafeteria and sports facilities) and existing contacts with training groups at the Olympic Center. Therefore, enrolled participants were physically active and recreational trained athletes, 18 to 50 years of age, of both genders (Tab. 10). 97 subjects were initially recruited for the study. After receiving an explanation of written informed consent protocols and additional oral information from the study coordinator, 94 (37m / 57f; 29±9yrs; 173±10cm; 71±13kg) subjects agreed to participate. All participants read and signed a written informed consent form before voluntary participation. The University´s Ethical Commission approved the study. With respect to the unequal distribution of subjects included in H and BPP, an additional matched-group analysis was conducted. Therefore, the equal number of healthy controls was gender-, age- and anthropometrically matched (group $H_{matched}$) to the number of back pain patients (BPP).

Measurement protocol

After receiving an anthropometric assessment, all subjects answered an online-based (ProWebDB, Germany) version of a back-pain questionnaire (von Korff) determining the presence of back pain [29]. The back pain questionnaire consisted of 7 items, including pain status and disability (recent and last 3 months)[29]. All items are conducted of a numeric rating scale ranging from 0 (no pain/disability) to 10 (highest pain/disability). In accordance with the grading score of the questionnaire, subjects were assigned to the healthy controls (H; Korff grades 0 and 1) or back pain patients group (BPP; Korff grades 2-4). Subjects were then prepared for EMG and kinematic analysis of the trunk. EMG electrodes were positioned over twelve trunk muscles (Fig. 16B; see EMG analysis section). Twelve reflective markers were precisely positioned over bony structures (Fig. 16A; see kinematic analysis section) [25]. Subject preparation was followed by a standardized walking perturbations protocol beginning with a warm-up and familiarization procedure where the participants walked 5 minutes at 1m/s on a split belt treadmill (Woodway, Weil am Rhein, Germany) without perturbation [25]. Next, each subject walked for about 10 minutes at a baseline velocity of 1m/s; while walking, 15 right- and left-sided perturbations were randomly applied 200ms after initial heel contact triggered by a plantar pressure insole (Pedar X, Novel, Munich, D). During perturbation, one of the treadmill belts decelerated by 2m/s resulting in a deceleration of 40m/s^2 for 50ms, returning to baseline velocity after an additional 50ms [25]. For the data analysis, only right-sided perturbations were analysed due to direct triggering of the perturbations by the plantar pressure insole used only in the right shoe. Left-sided perturbations were also applied to ensure that participants did not adapt their normal walking pattern to only right-sided perturbations.

Overall, subjects were commanded to walk as natural as possible on the treadmill while randomly perturbations will be applied. As a consequence, subjects walked on the treadmill while knowing that perturbations will be applied but not knowing when (time), where (leg) and how (direction). In addition, participants were instructed to compensate the stimuli aiming to get back to normal upright walking pattern within the following three to four steps (Cordero et al., 2003). No further instructions on arm, leg or trunk movements were given. For safety reasons, all participants worn a waist belt connected to an emergency stop release.

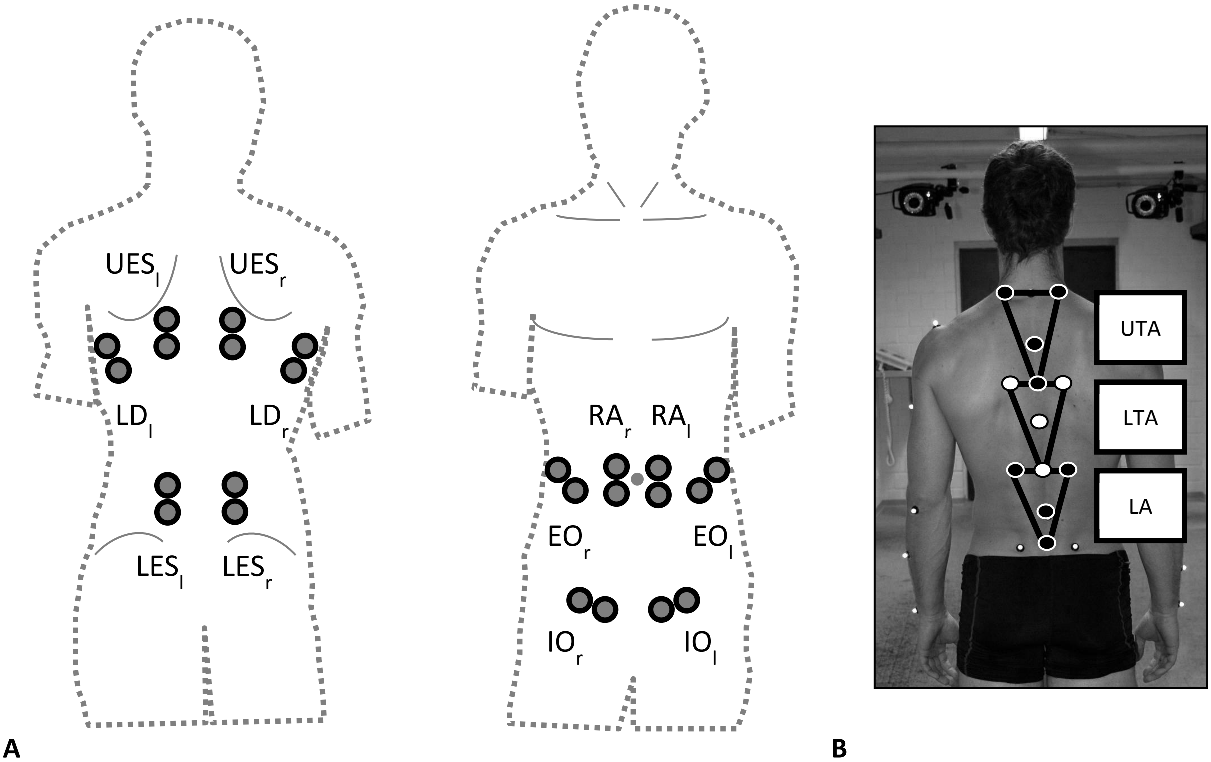
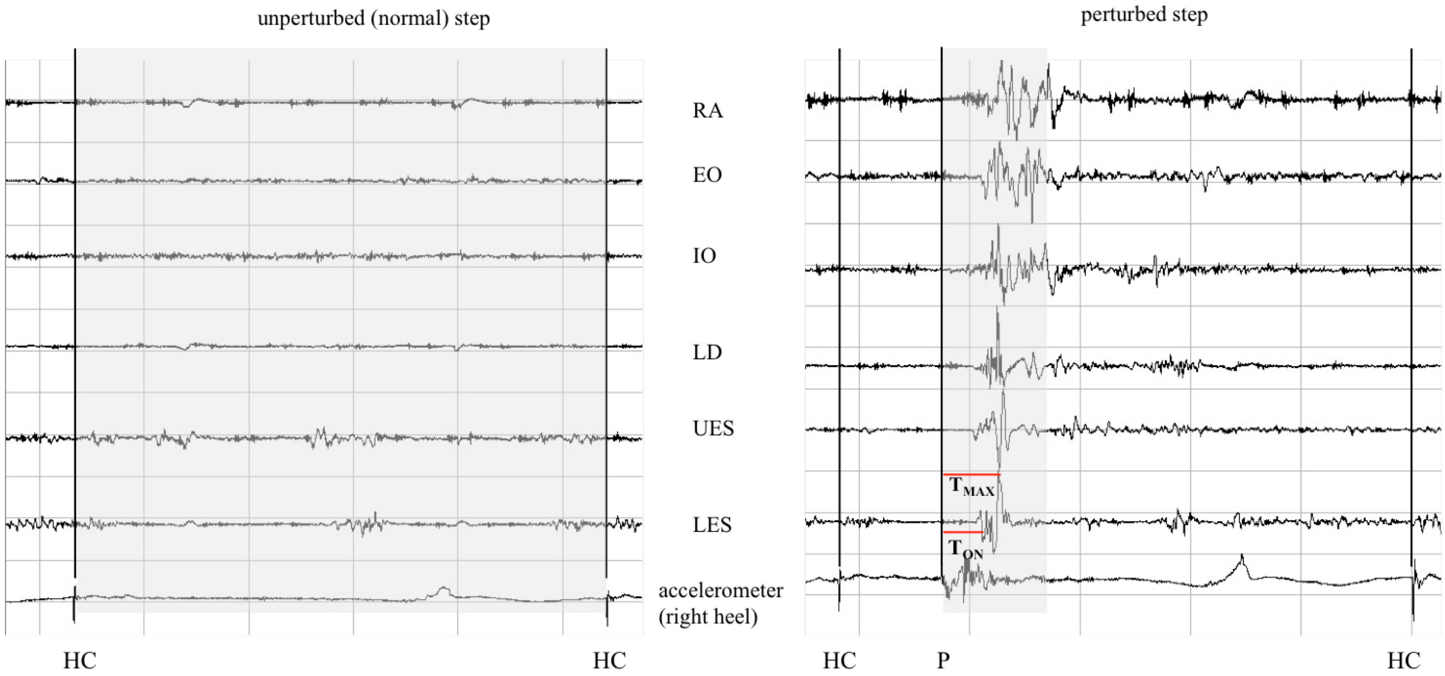


Figure 16: A. 12-lead EMG-trunk-setup; B. Kinematic trunk model (Müller et al. 2015)

EMG analysis

Trunk muscle activity was assessed with a 12-lead surface EMG (Fig. 16A) [17,30]. The setup included 6 ventral (Mm rec. abd. (RA), obl. ext. abd. (EO), obl. int. abd (IO) of left and right side) and 6 dorsal (Mm erec. spinae thoracic (T9; UES)/lumbar (L3; LES), latis. dorsi (LD) of left and right side) muscles. Muscular activity was analyzed using bilateral and bipolar surface EMG (bandpass filter: 5 – 500Hz; sampling frequency: 4000Hz, amplification: overall gain: 1000; myon, Switzerland). Before electrodes (AMBU Medicotest, Denmark, Type N-00-S, inter- electrode distance: 2 cm) were applied, the skin was shaved, slightly exfoliated to remove surface epithelial layers, and finally disinfected. In addition, skin resistance was controlled (<5kΩ). The longitudinal axes of the electrodes were in line with the presumed direction of the underlying muscle fibers. The mean amplitude for each muscle was calculated out of the first 5 unperturbed strides and the 15 perturbed strides of the walking perturbations protocol. As a main outcome measurement, the root mean square (RMS; [%]) normalized to the unperturbed stride was analyzed within the first 200ms following perturbation [31,32] (Fig. 2). Additionally, the mean (normalized) EMG-RMS for the right ventral area (VR: RA, EO, IO of right side), left ventral area (VL: RA, EO, IO of left side), right dorsal area (DR: UES, LES, LD of right side) and left dorsal area (DL: UES, LES, LD of left side) was build [23,30]. Therefore, the mean EMG-RMS of the three included muscles for each trunk area was calculated. The coefficient of variation (CV; EMG-RMS, %; formula: SD (15x EMG-RMS single muscle) / mean (15x EMG-RMS single muscle)) within the 15 perturbations served as the outcome measurement to account for the variability of trunk muscle activity between H and BPP.

In the time domain, the onset of muscular activity (T_{ON}; ms) and the time to maximum activity (T_{MAX}; ms) were measured, representing a response to the perturbation (Fig. 17). A semi-automated detection method (IMAGO process master, LabView®-based, pfitec, biomedical systems, Endingen, Germany) was used to define muscle activity onset [33]. Within this detection method, an increase in the averaged EMG signal (ensemble average; filter: 4th order moving average) of more than 2 standard deviations from baseline level was defined for automatic onset detection. Every automatic detection was controlled through visual inspection. If automatic detection failed (e.g. due to movement artefact), the investigator applied manual correction (<3% of all cases analysed).



Legend:
HC = heel contact
P = begin perturbation (200ms after HC)

= time window for RMS analysis (unperturbed: whole stride cycle; perturbed:200ms after perturbation)
T_{ON} = time to onset (ms)
T_{MAX}= time to maximum activity (ms)

Figure 17: Exemplarily EMG signal for the 6 right-sided trunk muscles (raw signal of 1 perturbation for one subject) including visualization of EMG outcome measures (EMG-RMS, T_{ON}, T_{MAX})

Kinematic analysis

Segmental trunk motion was measured using a 14-camera 3D-motion analysis system (Vicon, Oxford, UK, MX3, 1000Hz). The kinematic trunk model consisted of 12 markers framing three functional segments (upper thoracic area (UTA), lower thoracic area (LTA) and lumbar area (LA))(Fig. 16B)[25]. In addition, four markers framed the pelvis. Marker data were analyzed (Vicon Nexus 1.8) to calculate the relative angles of each segment in relation to the pelvis. The main outcome measurements were the total motion amplitudes (ROM; [°]) and mean trunk angle (A_{mean}; [°]). Both were calculated for normal (unperturbed) as well as perturbed gait for the whole stride cycle and the time interval 200ms following perturbation for extension/flexion (E/F), lateral flexion (LF) and rotation (Ro) of each segment. ROM consisted of the mean of the 15 repetitions following right-sided perturbations. Additionally, the coefficient of variation (CV; ROM, %, formula: SD (15x ROM of each segment per single plane) / mean (15x ROM of each segment per single plane)) within the 15 repeated perturbations was calculated for ROM. The mean trunk angle (A_{mean}; [°], mean of 15 repetitions) was calculated to describe the overall posture and its 3 segments over the whole stride cycle and 200ms after perturbation. By use of descriptive analysis (mean ± SD) the angle-time-curves for the LTA segment (in all planes) during walking and perturbation are presented as a group (H vs. BPP) and single case comparison (Fig. 20/21). With respect to the kinematic model used, negative values represent flexion, left-sided rotation and left-sided lateral flexion for all 3 segments. In contrast, positive values represent extension, right-sided rotation and right-sided lateral flexion.

Data analysis and statistics:

All non-digital data were documented in a paper and pencil-based case report form (CRF) and transferred to the statistical database (JMP Statistical Software Package 9, SAS Institute®). After a plausibility check (range check + extreme value analysis for all outcome measures), the data was analyzed descriptively (means, SD) for all given outcome measurements followed by a student´s t-test to account for differences between H and BPP. The level of significance was set at α=0.05. Multiple testing was controlled via Bonferroni adjustment (adjusted α =0.025)

8.4 Results

Back pain

80 subjects were allocated to the healthy control (H) and 14 to the back pain (BPP) group. This represents a back pain prevalence of 15% in this cohort. Anthropometrics and pain sub scores (pain intensity/disability score) of both groups are presented in Table 10. Statistically significant differences between H and BPP were present in the pain sub scores ($p<0.001$) but not in the anthropometrics. Regarding matched group analysis, 14 healthy controls ($H_{matched}$) were matched to the 14 back pain patients (BPP) and statistically significant differences between $H_{matched}$ and BPP were present in the pain sub scores ($p<0.001$) but again not in the anthropometrics. In addition, significant differences of acute back pain intensity at time of testing are present between H ($H_{matched}$) and BPP ($p=0.0001$ ($H_{matched}$: $p=0.003$)). H ($H_{matched}$) showed an intensity of 0.4±0.8 and BPP of 2.4±2.3 on a numeric rating scale ranging from 0 (no pain) to 10 (maximum pain).

Table 10: Anthropometrics and back pain status of healthy ($H/H_{matched}$) and back pain patients (BP) group

Group	N	Gender (f/m)	Age [yrs]	Body height [cm]	Body weight [Kg]	Korff Pain Intensity Score*	Korff Disability Score*
H	80	49/31	29 ± 9	174 ± 10	71 ± 13	17 ± 13	8 ± 11
BP	14	8/6	30 ± 8	171 ± 10	67 ± 14	50 ± 13	41 ± 18
$H_{matched}$	14	8/6	28 ± 8	170 ± 8	67 ± 12	13 ± 12	8 ± 11

* significant differences between H and BPP ($p<0.0001$)

\# significant differences between $H_{matched}$ and BPP ($p<0.0001$)

Trunk muscle activity following stumbling during gait

In the EMG-RMS analysis, no significant group differences (BPP vs. H; BPP vs. $H_{matched}$) of single muscles and muscle areas were found (p>0.025)(Tab. 11/Fig. 18). Variability of neuromuscular reflex activity, represented by CV, ranged from 23% to 37% for BPP, 25% to 39% for H, and 23% to 39% for $H_{matched}$ without significant differences between groups (p>0.025).

T_{ON} ranged from 78ms to 107ms in H, 74ms to 102ms in $H_{matched}$ and 82ms to 123ms in BPP (Tab. 11). BPP showed increased latencies for all 12 muscles compared to H, and were significant for RA le (p=0.019) and UES le (p=0.005). Matched group analysis also showed increased latencies for all 12 muscles in BPP compared to $H_{matched}$ and was statistically significant for RA le (p=0.004), UES le (p=0.005) and LES ri (p=0.004)(Tab. 11; Fig. 19A). For both group analyses, polar plots (Fig. 19A) show specific muscular reaction patterns for BPP and H/$H_{matched}$.

T_{MAX} showed increased values for BPP compared to H/$H_{matched}$ except M. latisimus dorsi left. Statistically significant differences between groups (BPP vs. H) could be observed for RA ri (p=0.021), EO ri (p=0.005) and LES ri (p=0.016). Matched group analysis (BPP vs. $H_{matched}$) only showed significant differences for RA left (p=0.004)(Tab. 11). The polar plot (Fig. 19B) specifies the different muscular reaction patterns of the groups.

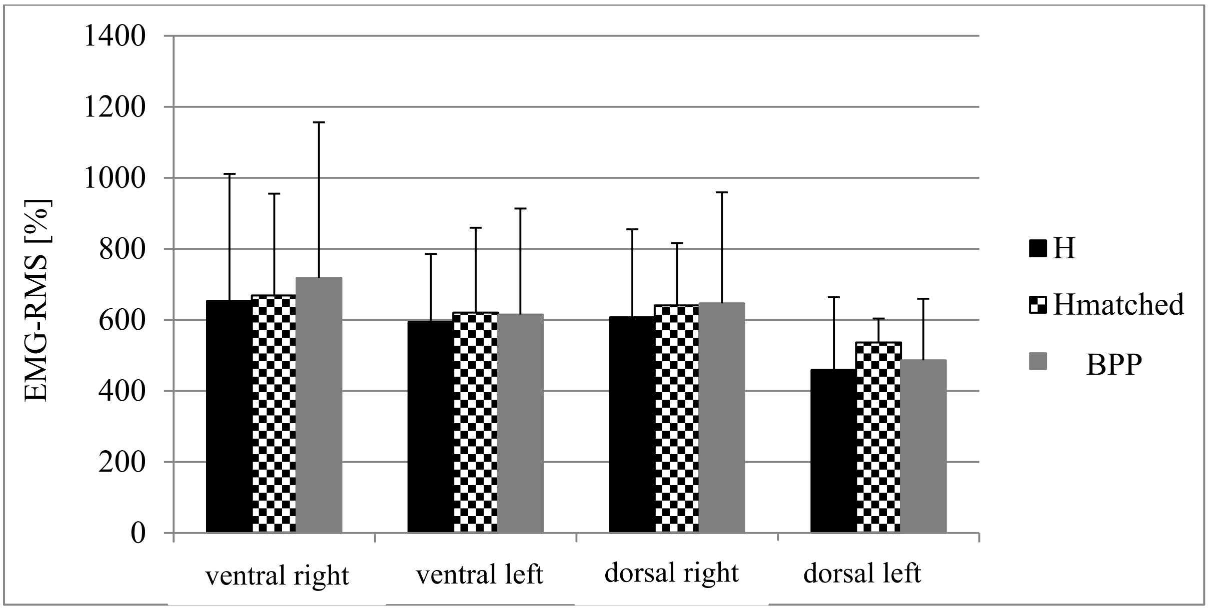


Figure 18: Neuromuscular reflex activity (EMG-RMS; %) of the four trunk areas during stumbling in healthy (H; $H_{matched}$) and back pain patients (BPP)

Legend: VR/VL: mean EMG-RMS of RA, EO, IO ri/le; DR/DL: mean EMG-RMS of LD, UES, LES ri/le)

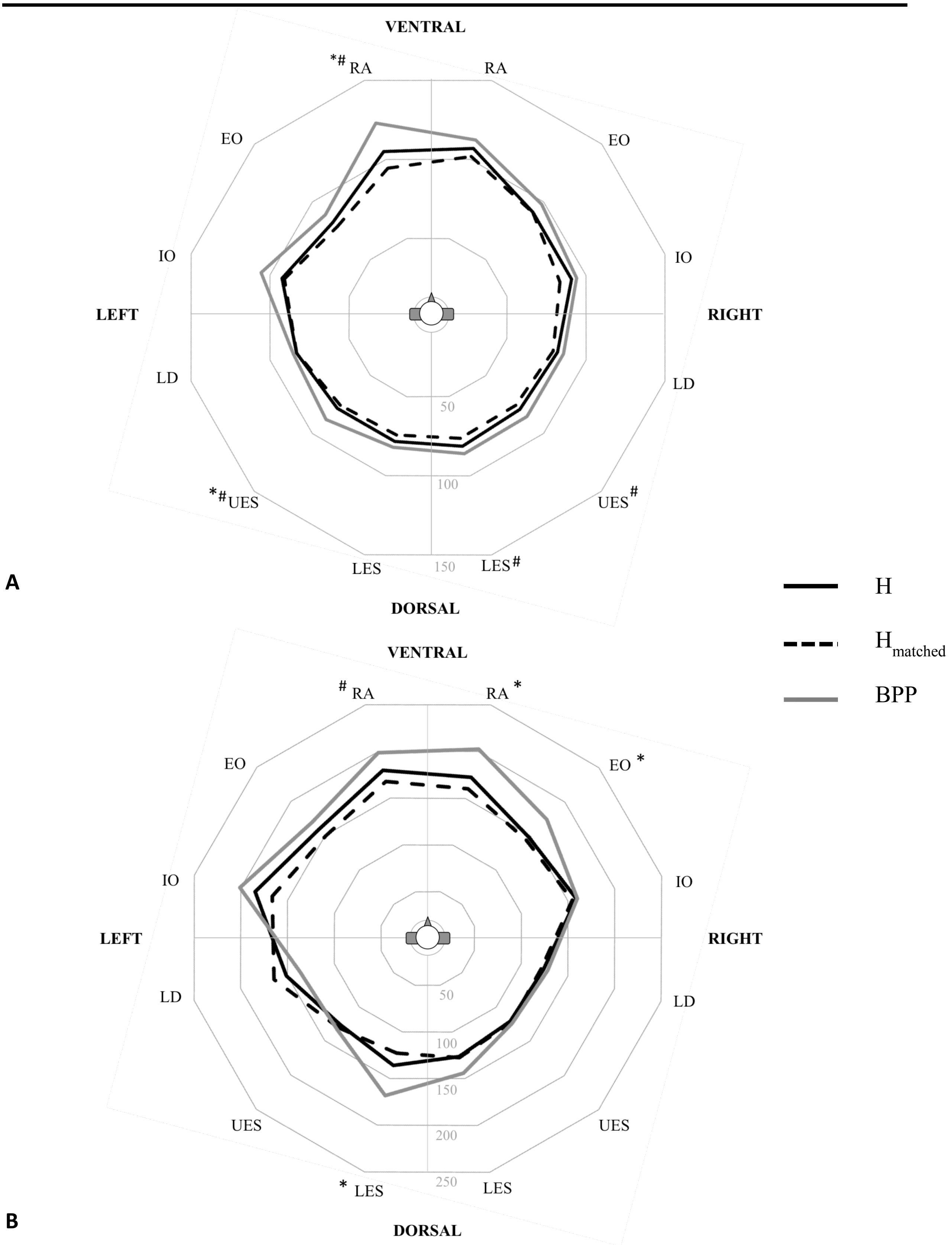


Figure 19: Polarplot of neuromuscular response time of 12 trunk muscles to perturbation

- **(A) onset of muscle activity (T_{ON}; ms) after perturbation**
- **(B) time to maximum activity (T_{MAX}; ms) after perturbation**

* significant differences between H and BPP ($p<0.025$); # significant differences between $H_{matched}$ and BPP ($p<0.025$)

Table 11: Neuromuscular reflex activity (amplitudes: EMG-RMS; %) and neuromuscular response (time measures: T_{ON}, ms; T_{MAX}, ms) for all muscles for BPP and H

outcome measure	groups	trunk muscles											
		RA ri	RA le	EO ri	EO le	IO ri	IO le	LD ri	LD le	UES ri	UES le	LES ri	LES le
Amplitudes													
EMG-RMS (%)	**H**	696 ± 712	602 ± 493	708 ± 407	670 ± 339	558 ± 314	513 ± 379	691 ± 403	546 ± 315	551 ± 231	367 ± 156	581 ± 271	465 ± 210
	$H_{matched}$	582 ± 199	576 ± 249	811 ± 536	764 ± 434	613 ± 461	522 ± 240	780 ± 310	699 ± 413	617 ± 246	434 ± 165	526 ± 170	476 ± 162
	BPP	649 ± 513	571 ± 306	873 ± 644	777 ± 280	633 ± 357	498 ± 232	808 ± 563	581 ± 280	505 ± 163	402 ± 323	554 ± 228	475 ± 246
	p-values BPP vs. H / BPP vs. $H_{matched}$	0.81 / 0.75	0.83 / 0.49	0.21 / 0.96	0.27 / 0.89	0.42 / 0.79	0.89 / 0.79	0.35 / 0.90	0.69 / 0.40	0.50 / 0.19	0.53 / 0.78	0.72 / 0.83	0.87 / 0.78
Time measures													
T_{ON} (ms)	**H**	107 ± 18	105 ± 22	91 ± 14	82 ± 12	91 ± 16	93 ± 18	82 ± 9	83 ± 10	79 ± 10	78 ± 10	81 ± 11	78 ± 9
	$H_{matched}$	102 ± 11	95 ± 13	91 ± 16	79 ± 11	84 ± 12	91 ± 19	79 ± 7	83 ± 10	76 ± 7	76 ± 7	77 ± 9	74 ± 8
	BPP	112 ± 21	123 ± 30*#	98 ± 15	89 ± 17	94 ± 20	106 ± 31	86 ± 10	85 ± 9	86 ± 9#	88 ± 12*#	86 ± 9#	82 ± 12
	p-values BPP vs. H / BPP vs. $H_{matched}$	0.43 / 0.14	0.019 / 0.004	0.09 / 0.24	0.13 / 0.10	0.56 / 0.14	0.04 / 0.16	0.22 / 0.07	0.50 / 0.55	0.046 / 0.005	0.005 / 0.004	0.16 / 0.015	0.24 / 0.07
T_{MAX} (ms)	**H**	173 ± 28	180 ± 31	148 ± 22	159 ± 41	159 ± 27	185 ± 37	124 ± 12	151 ± 38	121 ± 17	127 ± 31	126 ± 49	136 ± 91
	$H_{matched}$	160 ± 28	168 ± 25	144 ± 15	149 ± 27	155 ± 19	166 ± 25	121 ± 13	164 ± 53	122 ± 25	130 ± 35	128 ± 22	123 ± 29
	BPP	203 ± 78*	199 ± 23#	174 ± 56*	169 ± 25	160 ± 31	201 ± 71	128 ± 19	136 ± 23	124 ± 15	133 ± 31	144 ± 17*	169 ± 94
	p-values BPP vs. H / BPP vs. $H_{matched}$	0.021 / 0.07	0.06 / 0.004	0.005 / 0.06	0.42 / 0.72	0.94 / 0.62	0.22 / 0.10	0.29 / 0.30	0.24 / 0.14	0.60 / 0.86	0.58 / 0.84	0.017 / 0.26	0.27 / 0.10

* significant differences between H and BPP ($p<0.025$)
# significant differences between $H_{matched}$ and BPP ($p<0.025$)

Three-dimensional trunk kinematics during normal gait and following stumbling

Trunk motion analysis (ROM) over the whole stride cycle revealed no significant differences between groups ($p>0.025$) (Tab. 12). Analysis of reflex response (200ms after perturbation) showed a lower ROM (in relation to the pelvis segment) in compensation for perturbations for all 3 segments without statistically significant differences between groups (BPP vs. H; BPP vs. $H_{matched}$) ($p>0.025$) (Tab. 13). The CV for ROM ranged from 39±13 % (LA in lateral flexion) to 167±55 % (LTA in rotation) for BPP, 44±15 % (LA in lateral flexion) to 177±57 % (LTA in rotation) for H, and 43±13 % (LA in lateral flexion) to 182±60 % (UTA in lateral flexion) for $H_{matched}$. No group differences were found ($p>0.025$).

Regarding overall posture (A_{mean}) during normal gait, BPP showed a significant higher flexed posture of the UTA and LTA segment during normal walking (Tab. 12, Fig. 20/Fig. 21). However, no differences were found for the lumbar segment. Trunk posture analysis (A_{mean}) during reflex response showed higher trunk extension (E/F) values in LA and LTA segments for H/$H_{matched}$ compared to BPP (BPP vs H: $p=0.003$, BPP vs. $H_{matched}$: $p=0.015$; Fig. 22 / Tab. 13). Furthermore, BPP showed reduced lateral flexion in all 3 segments compared to H/$H_{matched}$ (Fig. 22) without significant differences ($p>0.025$). No differences were found for rotation ($p>0.025$) (Tab. 13).

Angle-time-crves, displayed in Fig. 20/21, visualize the above-described results for the LTA segment in a group (A) and single case (B) comparison.

Table 12: Three-dimensional trunk kinematics during normal, unperturbed and perturbed gait for the whole stride cycle (mean ± SD) for all three segments in all planes

(total motion amplitude (ROM, [°]); mean trunk angle (A_{mean}; [°]))

Segment	Plane	ROM (°)				A_{mean} (°)			
		BPP	**H**	**$H_{matched}$**	p-values BPP vs. H / BPP vs. $H_{matched}$	**BPP**	**H**	**$H_{matched}$**	p-values BPP vs. H / BPP vs. $H_{matched}$
Normal, unperturbed step									
UTA	**E/F**	9 ± 3	8 ± 3	7 ± 3	0.23 / 0.25	-17 ± 9*	-11 ± 7	-11 ± 6	0.02 / 0.08
	LF	11 ± 3	11 ± 3	10 ± 5	0.74 / 0.62	-1 ± 2*#	1 ± 2	1 ± 2	0.0006 / 0.020
	Ro	4 ± 2	4 ± 1	5 ± 2	0.83 / 0.86	2 ± 3*	-1 ± 3	-1 ± 3	0.008 / 0.09
LTA	**E/F**	4 ± 1	4 ± 1	5 ± 2	0.95 / 0.44	9 ± 7*#	15 ± 6	15 ± 6	0.004 / 0.02
	LF	9 ± 2	10 ± 3	10 ± 3	0.35 / 0.50	-1 ± 3	0 ± 2	0 ± 2	0.13 / 0.38
	Ro	10 ± 3	10 ± 3	10 ± 3	0.99 / 0.77	2 ± 4	1 ± 2	1 ± 3	0.38 / 0.50
LA	**E/F**	4 ± 1	4 ± 1	4 ± 1	0.65 / 0.37	3 ± 9	7 ± 6	6 ± 5	0.13 / 0.31
	LF	7 ± 3	6 ± 2	6 ± 2	0.23 / 0.23	-1 ± 3	0 ± 2	0 ± 1	0.22 / 0.50
	Ro	8 ± 3	8 ± 2	8 ± 2	0.73 / 0.87	1± 4	1 ± 2	1 ± 3	0.80 / 0.76
Perturbed step									
UTA	**E/F**	9 ± 2	9 ± 3	9 ± 3	0.84 / 0.69	-15 ± 9	-11 ± 7	-13 ± 5	0.08 / 0.42
	LF	14 ± 2	13 ± 4	13 ± 2	0.37 / 0.35	-1± 2*	2 ± 3	1 ± 3	0.001 / 0.16
	Ro	9 ± 2	9 ± 3	9 ± 2	0.79 / 0.66	2 ± 4	1 ± 3	1 ± 3	0.30 / 0.51
LTA	**E/F**	9 ± 2	9 ± 3	9 ± 3	0.58 / 0.79	9 ± 6*#	15 ± 6	15 ± 6	0.003 / 0.02
	LF	12 ± 3	12 ± 3	12 ± 3	0.73 / 0.71	-1 ± 3	1 ± 3	0 ± 3	0.06 / 0.38
	Ro	9 ± 2	10 ± 3	11 ± 3	0.28 / 0.04	2 ± 4	3 ± 3	2 ± 3	0.71 / 0.84
LA	**E/F**	7 ± 1	6 ± 2	7 ± 3	0.23 / 0.52	3 ± 9	6 ± 6	5 ± 5	0.24 / 0.39
	LF	8 ± 3	7 ± 3	7± 3	0.20 / 0.24	-1 ± 3	0 ± 2	0 ± 1	0.15 / 0.46
	Ro	9 ± 2	9 ± 2	9 ± 7	0.75 / 0.73	1 ± 5	2 ± 3	1 ± 3	0.47 / 0.94

* significant differences between H and BPP ($p<0.025$)
\# significant differences between $H_{matched}$ and BPP ($p<0.025$)

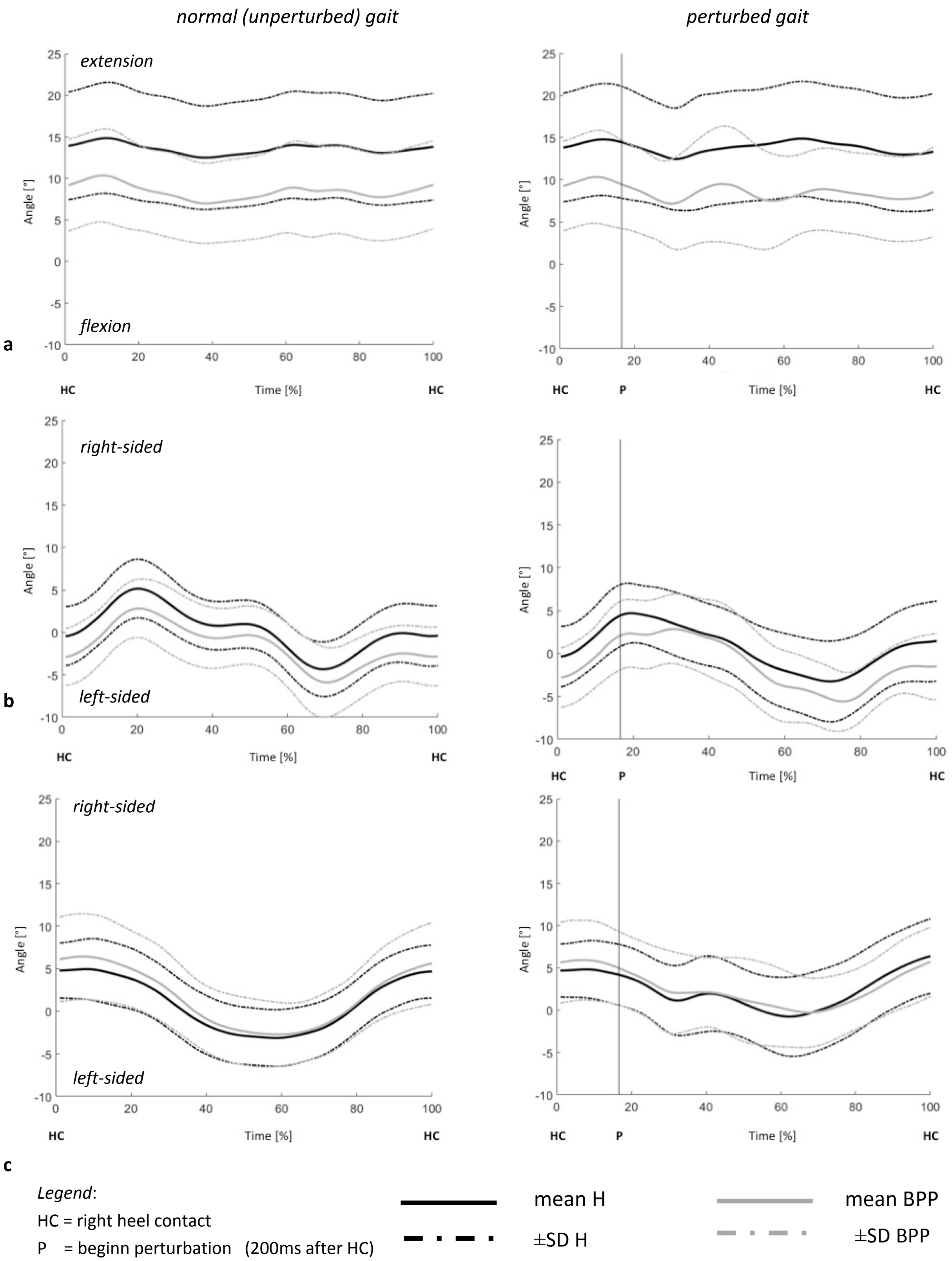


Figure 20: Group comparison of the LTA segment motion in (a) flexion, (b) lateral flexion and (c) rotation: Comparison of unperturbed and perturbed step in H and BPP (group mean ± SD)

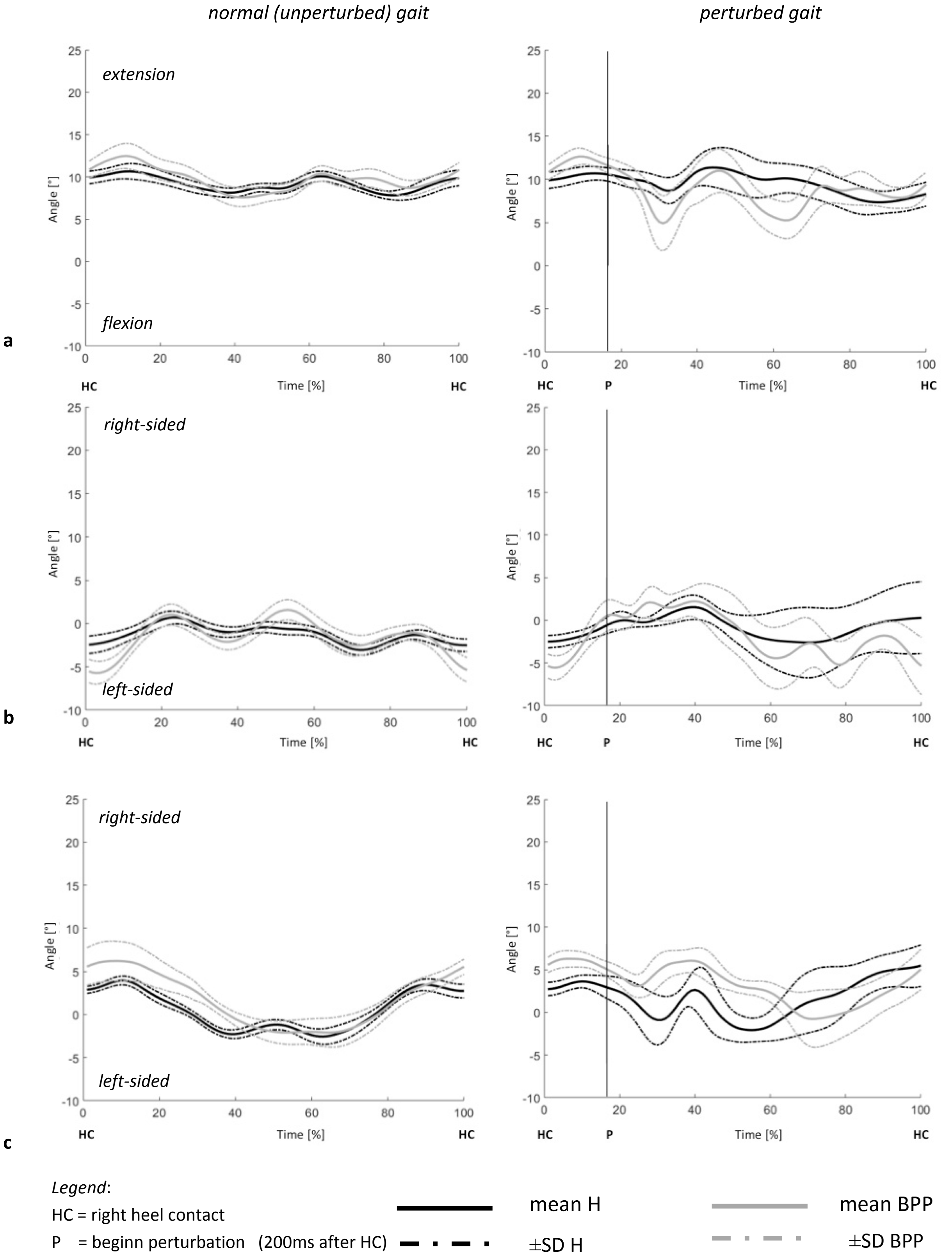


Figure 21: Case comparison of the LTA segment motion in (a) flexion, (b) lateral flexion and (c) rotation: Comparison of unperturbed and perturbed step in one healthy and back pain patient (individual mean ± SD of 15 perturbed and unperturbed steps)

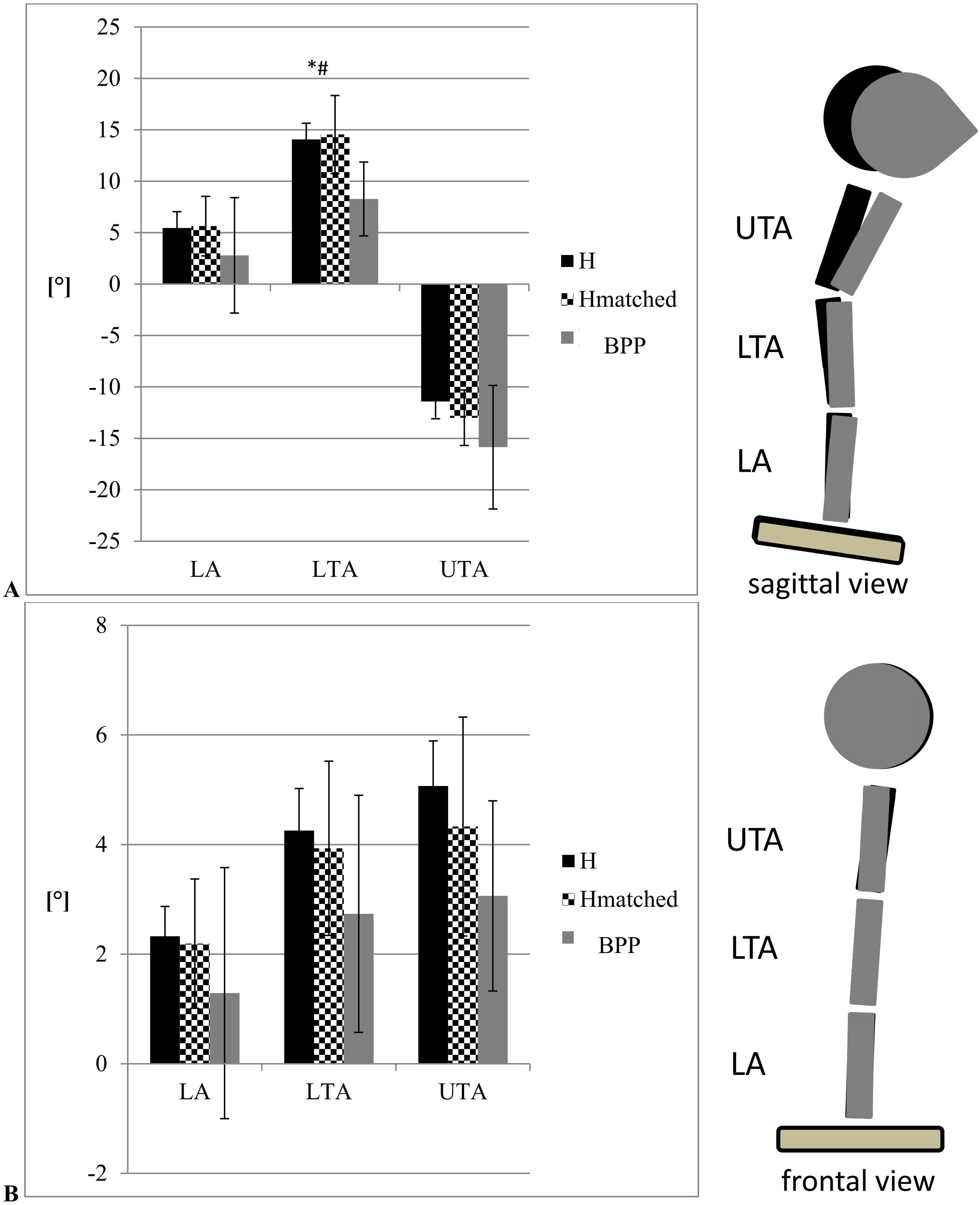


Figure 22: 3-D trunk kinematics during stumbling (mean trunk angle (A_{mean}; [°]; subsequent 200ms after perturbation) including mean ± 95%-confidence interval for (A) extension-flexion and (B) lateral flexion

* significant differences between H and BPP ($p<0.025$); # significant differences between $H_{matched}$ and BPP ($p<0.025$)

Table 13: Three-dimensional trunk kinematic reflex response during perturbation for the subsequent 200ms after perturbation (mean ± SD for all three segments in all planes)
(total motion amplitude (ROM, [°]); mean trunk angle (A_{mean}; [°])

Segment	Plane	ROM (°)				A_{mean} (°)			
		BPP	H	$H_{matched}$	p-values BPP vs. H / BPP vs. $H_{matched}$	BPP	H	$H_{matched}$	p-values BPP vs. H / BPP vs. $H_{matched}$
UTA	E/F	3 ± 2	3± 2	3 ± 2	0.97 / 0.93	-16 ± 10	-11 ± 7	-13 ± 5	0.05 / 0.34
	LF	4 ± 2	5 ± 2	5 ± 2	0.94 / 0.85	3 ± 2	5 ± 3	4 ± 3	0.05 / 0.31
	Ro	3 ± 2	3 ± 1	3 ± 1	0.95 / 0.29	1 ± 5	-1 ± 3	0 ± 2	0.19 / 0.52
LTA	E/F	4 ± 2	3 ± 2	3 ± 2	0.46 / 0.47	8 ± 6*#	14 ± 6	15 ± 6	0.003 / 0.015
	LF	3 ± 1	4 ± 2	4 ± 2	0.15 / 0.13	3 ± 3	4 ± 3	4 ± 3	0.12 / 0.33
	Ro	5 ± 2	4 ± 2	4 ± 2	0.69 / 0.40	3 ± 4	3 ± 3	3± 3	0.89 / 0.98
LA	E/F	4± 1	4 ± 2	4 ± 2	0.68 / 0.54	3 ± 9	5 ± 6	6 ± 5	0.21 / 0.32
	LF	2 ± 1	2 ± 1	2 ± 1	0.57 / 0.62	1 ± 4	2 ± 2	2 ± 2	0.18 / 0.44
	Ro	4 ± 1	4 ± 1	4 ± 1	0.82 / 0.68	0 ± 5	1 ± 3	1± 3	0.64 / 0.75

* significant differences between H and BPP ($p<0.025$)
\# significant differences between $H_{matched}$ and BPP ($p<0.025$)

8.5 Discussion

The main purpose of this study was to analyse trunk stability during sudden external loading while walking, characterized by EMG reflex activity and three-dimensional segmental motion of the trunk in back pain patients (BPP). This study demonstrates that BPP showed a significantly later onset of some muscle responses to sudden loading without differences in the neuromuscular reflex amplitude. Furthermore, significant differences in patients' overall trunk posture during normal gait and following stumbling were found compared to healthy controls.

A delayed trunk muscle response to sudden force release in back pain patients is in line with other studies [9,17]. In addition, a prolonged muscle response until maximum activity was found in patients, which has not been previously described elsewhere. Stumbling while walking represents a functional and daily-life situation in which sudden unexpected loading increases the risk of overloading and injury (e.g. slipping or tripping while walking). The influence of back pain on neuromuscular responses of the trunk to sudden loading could be shown. The majority of studies so far have mostly evaluated more static, seated or standing positions, aiming at isolated trunk muscle activity analysis [9,16]. Furthermore, conclusions regarding neuromuscular deficits have mostly been attributed to direct loading of the trunk muscles and have not involved load applied through the extremities such as during gait [9,17,30,34]. The presented results support exercise interventions addressing the trunk not only directly but also indirectly by distal segments (upper or lower extremities) [35].

Detailed neuromuscular response pattern analysis (T_{ON}; T_{MAX}) could show that patients in particular alter their reaction time of bilateral trunk muscles when the right side is perturbed. Therefore, an impaired neuromuscular control of the trunk has to be assumed. This, in particular, might also have implications for training strategies taking one-sided perturbations into account to address bilateral neuromuscular core muscle stability.

No differences in neuromuscular reflex amplitudes could be shown between patients and healthy controls. Lamoth et al. (2005) [36] reported impaired muscle coordination and increased muscle activity of the erector spinae (ES) in low back pain patients compared to healthy subjects during normal, unperturbed walking at different velocities. The results of the present study imply an impaired trunk muscle response to sudden loading in back pain patients while walking. Additional alterations or restrictions on the magnitude of neuromuscular reflex activity in patients could not be observed. These might be interpreted as a quick-targeted neuromuscular response of healthy subjects to sudden walking perturbations. In contrast, delayed BPP response could be caused by pain-inhibited proprioception. Similar EMG amplitudes between groups could stand for an overshooting action to compensate delayed activation onset, despite reduced activation time in the analysed time window.

As mentioned above, kinematic outcomes during walking and back pain are diverse [24,26,27,37]. Vogt et al. (2001) [27] demonstrated that lumbar trunk motion patterns and displacement in patients while walking were equivalent to those of healthy controls. In contrast, the presented results add, that during normal walking as well as stumbling BP patients show an altered kinematic trunk motion pattern with significant differences, especially for the upper and lower thoracic segment. As shown in this study, these differences are persistent during reflex response to walking perturbations representing a characteristic compensation pattern with counter movements predominantly in sagittal plane. Moreover, significant differences in overall trunk posture (A_{mean}) after stumbling in BPP compared to H could be observed for the transversal and sagittal planes. In addition, patients showed a more flexed posture of the trunk, especially in the upper thoracic region. This might be discussed as a relieving or protective posture due to pain or fear of pain involved in walking and repetitive external gait perturbations. From a biomechanical perspective, higher flexion angles of the trunk imply greater lever arms generating higher moments. This alteration could indicate reduced trunk stability and higher loading during gait perturbations in BPP.

In BPP muscles response times and overall trunk posture, especially in the sagittal and transverse planes, are impaired. As a consequence, it might be speculated that the back pain patients were not able to immediately and adequately compensate for repetitive, sudden loading while walking. Hence, BP patients are at potentially higher risk of overloading and injury when exposed to repetitive external loading (e.g. slipping or tripping) [9].

The recently discussed higher variability of muscular activity in patients compared to healthy controls could not be supported by the results presented here [24,36]. Both groups showed high variability in intra-individual neuromuscular reflexes, as calculated by CV over 15 repetitions, in response to the unexpected, high-intensity loading situation [25]. In contrast to others, no differences in the movement variability of inter-segmental trunk motion (ROM) could be demonstrated in BPP [27]. Even though the variability is high in both groups, the absolute ROM is low. Asgari et al. [14] similarly reported no differences in trunk motion variability between low back pain and healthy controls in a flexion-extension task. They demonstrated that a higher movement speed significantly reduced trunk kinematic variability in both groups. With respect to Asgari et al. [14], the high intensity of the chosen perturbation combined with the low overall ROM of the trunk might be responsible for the small differences between healthy controls and back pain patients in the present study [25].

In summary, back pain patients present an altered neuromuscular and kinematic compensation strategy in response to unexpected sudden loading while walking compared to healthy controls. The delay in muscle activity combined with an altered kinematic trunk posture might lead to higher loading and reduced overall stability of the trunk. Therefore, it could be speculated, that exercise therapy in prevention and rehabilitation of back pain should include various perturbations with sudden, unexpected loading strategies in subjects that are at risk of developing back pain due to unexpected trunk loading [35]. Sensorimotor training (SMT), as described in previous studies including additional external perturbations, seems to be a feasible option for enhancing performance of the trunk muscles [4,38,39]. Further validation of this approach is required by randomized controlled trials.

Certain limitations have to be considered when interpreting the results. During the experiment, all subjects walked at the same baseline velocity, not taking into account a potentially reduced self-selected (comfortable) gait velocity in back pain patients [36]. With respect to standardization, a consistent test situation for all subjects was favored anyway [40].

For the data analysis, only right-sided perturbations were analysed. It cannot be ruled out that participants were stressed to different extends due to individual foot dominance. Nevertheless, the human gait is described as an automated and stable movement pattern (high intra-individual reproducibility). Consequently, there is no need to expect asymmetries in participants without pain, complaints and/or injuries at the lower limbs that was ensured by a clinical examination conducted by an experienced physician. Besides, no specific instructions regarding the task of the trunk, legs or arms during compensation were given to the participants. Use of different compensation strategies (e.g. leg-dominant, trunk-dominant) was not assessed. Different strategies might have influenced the presented neuromuscular and kinematic trunk response pattern of both groups to different amount.

Except for the sample size, there were no baseline (anthropometric) differences between groups. The added matched group analysis (BPP vs. $H_{matched}$) did not change the results of trunk EMG and kinematics. Since matching can reduce cofounding factors, it can also eliminate possibly important influencing effects. Therefore, necessity of matching in a cohort of adult subjects has to be discussed.

Conclusion

Back pain patients demonstrate different neuromuscular and kinematic compensation strategies for sudden loading while walking, presenting increased latencies in muscle reaction without differences in neuromuscular reflex amplitudes. In addition, overall trunk posture, especially in the sagittal and transversal planes, is altered in BPP. This might be discussed as relieving posture during normal walking and persistent during provoked stumbling. Accordingly, exercise therapy might aim for the improvement of trunk muscle response to sudden unexpected perturbations during dynamic tasks and overall trunk posture during walking. Sensorimotor training in combination with perturbation seems to be suitable. Nevertheless, future research is needed to validate this approach.

ACKNOWLEDGEMENTS

*The present study was initiated and funded by the German Federal Institute of Sport Science and realized under the auspices of MiSpEx – the National Research Network for Medicine in Spine Exercise (grant number: BISp IIA1-080102A/11-14). The present study was also funded by the European Union (ERDF – European Regional Development Fund; grant number: 80132471).

The authors thank Martin Löhrer for writing the Bodybuilder Script. They are grateful to Stephan Kopinski, Konstantina Intziegianni, Hannes Kaplick, Gerome Vivar and Monique Wochatz for assisting with measurements and data analysis.

CONFLICTS OF INTEREST

There is no conflict of interest.

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9. Main Findings

The aim of the present thesis was to analyse the effect of back pain on 3D trunk kinematics and neuromuscular (reflex) response to continuous and sudden loading in everyday situations. Therefore, a reliable and valid measurement set-up was developed to assess trunk motion and neuromuscular acitivty in highly dynamic, everyday loading situations generated by the lower or upper limbs.

Main findings of the project “Trunk loading and Back pain”:

What does the project add?

- Trunk motion can reliably be measured using the newly developed multi-segmental trunk model during sudden and continuous trunk loading (lifting/gait/stumbling). Restrictions while analysing the lateral flexion have to be taken into account.
- The results support the need for a multi-segmental model to examine trunk motion in continuous (lifting) und sudden (walking perturbations) trunk loading.
- Sudden trunk loading (perturbed walking) leads to an increased trunk ROM and neuromuscular activity of the trunk muscles in healthy subjects. In contrast, continuous loading (lifting) with different weights did not signficiantly influence the multi-segmental motion pattern.

- Back pain patients demonstrate altered neuromuscular and kinematic compensation strategies of the trunk in response to sudden, unexpected walking perturbations compared to healthy controls.
    - Latency in response to sudden walking perturbations is significantly increased in BPP compared to healthy controls.
    - BP patients show an altered kinematic trunk motion pattern with significant differences, especially for the upper and lower thoracic segment and the sagittal and transverse planes.
    - BPP show a significantly more flexed posture of the trunk (upper thoracic area) compared to H during normal gait and compensation of perturbations.

In addition, the general results of the different studies are presented below according to the research questions (F1-F3).

Research Question 1:

Reliability of trunk muscle activity and kinematics

- Kinematic variables can be assessed, but the smaller the motion amplitude (ROM), the lower the reliability of the outcomes.
- The activity of the trunk muscles can be reliably measured by means of a 12-lead EMG set-up during dynamic loading situations (lifting/gait/stumbling).
- The grouping of muscles is a methodological means to improve reliability of the outcome measures.

Validity of trunk muscle activity and kinematics

- The lifting of loads highly corresponds to a mundane motion task of the trunk with necessary motion amplitudes (ROM) in all segments of the trunk model (in all motion planes).
- The normal gait highly corresponds to a standardised motion task with an automated movement pattern. Application of additional perturbations during normal gait leads to increased coordinative requirements that are quantifiable with a measurable displacement of the trunk as well as the neuromuscular activity of the trunk muscles.
- Artificially stumbling, regardless of perturbation characteristics, leads to an increase in neuromuscular activity of the trunk's encompassing muscles in comparison to normal, unperturbed gait.
- The magnitude of the amplitude of the perturbation significantly influences the neuromuscular trunk response in healthy adults significantly.
- The neuromuscular activity pattern reflects both the intensity and localisation of the perturbation: an increased activity of the trunk muscles ipsilateral to the localisation of the perturbation.
- The response times (latencies) of the trunk muscles to different gait perturbations can be identified as a polysynaptic reflex response (medium latencies).
- The neuromuscular response times differ significantly between four different perturbations for dorsal, but not ventral muscles.

Research Question 2_a:

- The ROM differs significantly between the three trunk segments used, regardless of weight.
- There are no differences in segmental ROM between the two weights analyzed (1 Kg vs. 10 Kg). During lifting, the applied loads can be compensated by healthy individuals. Higher loads seem to be necessary to overreach the individual kinematic and neuromuscular margins of trunk stability.
- There are no interaction effects between the two selected loads and the three trunk segments.
- Continuous loading (lifting) of different weights over a defined period requires no additional adaptation of the kinematic trunk motion in healthy subjects.

Research Question 2_b:

- Artificially induced (one-legged) stumbling leads to a specific muscular compensation pattern (without statistically significant differences) in asymptomatic individuals:
    - increased activity of the ventral compared to the dorsal muscles
    - increased activity of the muscles on the side of the localisation of the perturbation in comparison to the unperturbed side.
- Artificial (one-legged) stumbling leads to a specific kinematic compensation pattern of the trunk: increased (left-sided) lateral flexion in combination with an increased (left-sided) rotation and flexion.
- Artificially induced (one-legged) stumbling leads to significant alterations in the ROM (for all 3 segments), but only for lateral flexion in comparison to normal gait.

Research Question 3:

- Back pain patients show a significantly increased time to onset and time to maximum activity of the trunk muscles in response to sudden walking perturbations in comparison to healthy controls.
- There are no significant differences in the amplitude of the neuromuscular reflex activity between healthy and back pain patients in response to sudden walking perturbations.
- Significant differences in back pain patients´ overall trunk posture during normal gait and following stumbling were found compared to healthy controls, especially in the sagittal and transverse planes.
- BPP show a significantly more flexed posture of the upper thoracic area compared to H during normal gait and compensation of perturbations.
- There are no differences in the maximum ROM (all 3 segments) of the trunk in healthy controls and BP patients in response to walking perturbations.

10. Discussion

10.1 Assessing trunk stability

10.1.1 Definition of a measurement set-up

The results from the reproducibility analysis support the clinical applicability of the applied methods (kinematics; EMG) and loading situations. The assessment of trunk motion shows acceptable to excellent reliability during dynamic continuous and sudden loading. Besides this, the analysis of the lateral flexion, assessed with the developed 3-segmental marker set-up, has to be discussed critically. Reliability of the trunk lateral flexion measurements has been described as reduced in previous investigations (Bible et al. 2010; Brink et al. 2011; Leardini et al. 2011; Schinkel-Ivy et al. 2015; Bauer et al. 2015a; Van Daele et al. 2007). Schinkel-Ivy et al. (2015) concluded that, particularly for motions with a small ROM, reduced reliability values are frequent. This corresponds with the presented results of the lateral flexion of the newly developed trunk segments during sudden and continuous trunk loading. It can be speculated that lateral flexion therefore guarantees functional variability for the control and regulation of a straightforward primary movement direction of the trunk, to ensure stability while compensating additional loading. In contrast, the primary movement direction (flexion/extension) as well as the rotation motion represent a stable pattern, while lifting loads and walking with and without perturbation (Dietz et al. 2002; Duysens & Van de Crommert 1998; Todorov and Jordan 2002). Todorov & Jordan (2002) discuss this observation as an optimised movement strategy that uses increased motion variability in the less relevant secondary movement directions (lateral flexion; rotation), aiming to optimize the motor control of a segment.

The results of the neuromuscular activity analysis show acceptable to insufficient reliability results in dynamic loading situations. This can be observed for the lifting task with respect to the high degree of freedom of the trunk (no fixation of body segments) during the performance of the tasks. In contrast, the amount of the increase in the (normalized) EMG amplitude for the targeted stumbling task allows an interpretation of the results despite the limited reproducibility (e.g., 40 % test-retest variability vs. up to 600 % increased EMG-RMS). The magnitude of the changed activation is therefore deductive.

The examined loading tasks are to be judged as highly relevant in characterising trunk motion in everyday movements, even though different demands on trunk stability could be observed between the two tasks (Maaswinkel et al. 2016; McGill et al. 2013). Lifting different loads, an expected and continuous loading task (continuous loading), leads to a noticeably lower displacement of the trunk or alteration of the neuromuscular activity in comparison to stumbling, an unexpected, sudden loading event (Müller et al. 2016a; 2016b). The implementation of these sudden perturbations leads to greatly increased requirements on the stability of the trunk (neuromuscular, kinematic). Maaswinkel et al. (2016) discuss the need to integrate perturbations into a diagnostic tool aiming to assess trunk stability. Furthermore, the authors claim to prefer a local perturbation applied directly to the trunk instead of a perturbation of the limbs. This is based on the difficulty of quantifying the real magnitude of the perturbation at the trunk as well as the magnitude of the compensation of the perturbation by the limbs, joints or other body segments (Maaswinkel et al. 2016). Nevertheless, the results of this thesis show that the dynamic loading situations used are suitable to cause and assess a trunk adaptation and compensation reaction in healthy participants. The participants' trunk stability, with their individual anthropometrics (e.g., height, weight, walking speed, step length), might be influenced to different extents by the applied walking perturbation. Still, it must be mentioned that both measurement methods are able to differentiate between healthy and BP patients, especially during perturbed walking. Based on the thesis results, a high degree of functionality and its link to everyday movement tasks is suitable to assess and evaluate trunk stability (Müller et al. 2016a; 2016b).

Consequently, a set-up to measure and evaluate trunk stability in healthy and back pain patients should include the following components based on the thesis outcomes:

(1) Loading tasks:

- Highly dynamic, daylike tasks with suddenly and unexpected applied loading (perturbations) that can be generated by the limbs (e.g., walking with perturbation),

(2) Measurement methods:

- Assessment of the neuromuscular response to a perturbation (e.g., EMG amplitudes, muscle response times)
- Assessment of the kinematic multisegmental trunk response to a perturbation (e.g., ROM, posture).

10.1.2 Effects of sudden and continuous loading on trunk stability

The applied loading situations represent highly dynamic, mundane motion tasks in which loading is initiated by the limbs (Burgess et al. 2009; McGill et al. 2013; Müller et al. 2016). Nevertheless, a comparison of the influence or magnitude of the different types of strain show different results with regards to trunk stability (Müller et al. 2016a; 2016b).

The sudden loading of the trunk by the applied perturbation of the leg caused a measurable alteration of the trunk as well as neuromuscular activity in healthy individuals (Müller et al. 2016). Therefore, the elective loading situation seems suitable to increase the demand for stabilisation of the trunk (Grabiner et al. 1993; Tang et al. 1998). An increased segmental lateral flexion and a

pronouncedly increased neuromuscular reflex activity of the ventral trunk muscles can be interpreted as a specific compensation pattern in asymptomatic adults.
In contrast, continuous loading of the trunk during lifting of different weights could not cause evident kinematic alterations in healthy subjects. Since lifting movements are omnipresent in everyday life, they can be described as automated movement patterns (Dietz et al. 2002; Duysens & Van de Crommert 1998). This might serve as an explanatory model why no kinematic changes of the trunk motion could be assessed with higher weight. On the other hand, McGill et al. (2013) have demonstrated that lifting heavier weights leads to higher spinal loading while carrying loads. Consequently, the absence of kinematic differences between light and heavy weights can be discussed as a result of adapted neuromuscular activity patterns of the trunk muscles in the cohort examined (Watanabe et al. 2012; Yoon et al. 2012). Therefore, it can be concluded that healthy subjects stabilise their trunk through an active neuromuscular compensation strategy involving automated movements during continuous loading (Watanabe et al. 2012). An additional increase in the weight used seems necessary to overcome individual margins of kinematic and neuromuscular trunk stability. The individual state of "neuromuscular failure" must be reached, e.g., by overcoming individual fatigue limits, in order to probably be able to detect kinematic alterations. The asymptomatic participants were able to compensate the additional weight during the automated lifting tasks. Consequently, the loading situation used in its current format must be discussed critically in terms of assessing trunk stability.

As a result, the loading situations analysed allow evaluating trunk stability in functional, everyday movement tasks. The evident differences between the loading tasks lead to the conclusion that the application of highly dynamic loading situations including perturbations should be preferred for trunk stability analysis in healthy and BPP.

10.1.3 Methodological Considerations

The application of optoelectronic (motion) measuring systems is described as the gold standard for non-invasive analysis of the functional trunk motion (Bauer et al. 2015b; McGinley et al. 2009). Thus, a three-dimensional (camera-based) motion measuring system (Vicon, UK) was preferred. Nevertheless, the application of such a measuring system in clinical practise is criticized due to high costs, special demands for room settings and staff, high demands for standardization of the marker positioning, as well as the data analysis and processing. Therefore, alternative measurement systems are necessary for the examination of the trunk's motion amplitude (ROM) and motion control in the clinical routine (Laird et al. 2014). Bauer et al. (Bauer et al. 2015a) examined the reliability of a wireless measurement device consisting of an inertial motion unit (IMU) including a sensor cluster that could measure motion angles as well as the speed of or between body segments based on the magnetic field and the gravitation (Roetenberg et al. 2007). The IMU motion measurement systems allow the reliable and valid assessment of the trunk motion in the primary motion plane (flexion/extension). In contrast, the analysis of the secondary motion plane (rotation, lateral flexion) showed reduced reliability (Bauer et al. 2015a). Nevertheless, Bauer et al. (2015b) demonstrate an influence of back pain on the lumbar motion patterns in a comparison to healthy based on the IMU system. For clinical application with the aim of a standardised, objective (simple) motion analysis based on the ROM, the axis problems are to be judged as less problematic. In contrast, for its application in the field of scientific research, this has to be discussed critically.

10.2 Trunk loading and back pain

10.2.1 Effects of back pain on trunk stability

The presented thesis could prove that BP patients show an altered neuromuscular and kinematic compensation pattern in response to an external perturbation during normal gait compared to healthy adults. A delayed response time and extended duration until the achievement of maximum activity in combination with a more flexed posture of the trunk, especially in the upper thoracic region, during perturbation is present. From a biomechanical perspective, higher flexion angle of the trunk imply greater lever arms generating higher moments, especially during repetitive loading of the trunk. These alterations could indicate reduced trunk stability and higher loading during gait perturbations in BPP. The results are in line with previous studies analysing trunk stability during perturbation (Cholewicki et al. 2000; Maaswinkel et al. 2016; Radebold et al. 2000). Conversely, the presented results gained by means of functional, dynamic loading tasks could confirm the findings from the quasi-static experiments (Cholewicki et al. 2000; Radebold et al. 2000). Moreover, dynamic loading tasks put more complicated sensorimotor and central nervous demands on the regulation mechanisms and thus allow for a complex examination of an individual's compensation abilities. The quasi-static experiments allow for a detailed evaluation of postural control. In contrast, the dynamic loading situations allow for analysis of the postural control in combination with motion pattern analysis.

The stratification of the examined variables highlights that, especially in the analysis of the neuromuscular response time and overall posture, a differentiation between healthy and back pain patients is successful (Granata et al. 2001; Mannion et al. 2012; Maaswinkel et al. 2016; Radebold et al. 2000). The above-presented response times of the trunk muscles ranged from 78±10ms to 107±18ms in healthy controls and from 82±12ms to 123±30ms in BPP. In sum, the controls showed reactions in the area of medium latency with polysynaptic reflex activity (50-80ms) and trigger reactions (80-120ms) (Milosevic et al. 2016). On the other hand, BP patients showed triggers (80-120ms) as well as voluntary reactions (120-180ms) (Milosevic et al. 2016). Besides, it has to be considered that mechanical delay from the distal applied perturbation to the trunk segment might cover these response times. Maaswinkel et al. (2016) discussed that an increased latency in combination with an increased co-contraction of the trunk muscles in BP patients can be interpreted as a functional adaptation to enhance trunk stability. Differences in the neuromuscular reflex activity (co-contraction) of the trunk muscles could not be shown during the stumbling experiment. Therefore, this interpretation by Maaswinkel et al. (2016) (especially in combination with the extended duration time before achieving maximum activity) could not be validly used for the stumbling reactions. In addition, these alterations might be interpreted as a quick-targeted neuromuscular response of healthy subjects to sudden walking perturbations. In contrast, delayed response in BPP could be caused by pain-inhibited proprioception. Similar EMG amplitudes between groups could stand for an overshooting action to compensate delayed activation onset, despite reduced activation time in the analysed time window.

During normal walking as well as stumbling BP patients show an altered kinematic trunk motion pattern with significant differences, especially for the upper and lower thoracic segment. These differences are persistent during reflex response to walking perturbations representing a characteristic compensation pattern with counter movements predominantly in sagittal plane. Moreover, significant differences in overall trunk posture after stumbling could be observed in BPP for the transverse and sagittal planes. Moreover, patients showed an increasingly flexed posture during normal gait and walking with perturbation, particularly in the upper thoracic area (UTA). This might be seen as a relieving or protective posture due to pain or fear of pain involved in walking and repetitive external gait perturbations. From a biomechanical perspective, higher flexion angles of the trunk imply greater lever arms generating higher moments. This alteration could indicate reduced trunk stability and higher loading during gait perturbations in BPP. Differences in the ROM or the variability of the motion, known from previous studies, could not be confirmed by this experiment (Asgari et al. 2015; Lamoth et al. 2005; Steele et al. 2014; Vogt et al. 2001;). The location of the applied perturbation (leg) might be discussed as a possible cause leading to only small displacements of the trunk's ROM (Maaswinkel et al. 2016).

The increased response time of the muscles combined with an altered (kinematic) posture of the trunk can be discussed as causing a changed and possibly increased loading of the spine, resulting in reduced trunk stability. Following these results, BP patients seem to have an increased risk of overuse and injury at the trunk in highly dynamic, repetitive loading situations including sudden, unexpected perturbations (e.g., stumbling).

10.2.2 Implications for prevention and therapy of back pain

The results of the present thesis have pointed out the need for BP prevention and therapy to focus on stabilising the trunk (Mannion et al. 2012; Pedersen et al. 2004; Searle et al. 2015; Saragiotto et al. 2016). The active compensation of external loading by the trunk is therefore of great relevance (Müller et al. 2016a; 2016b). Improving the neuromuscular control of the trunk has to be seen as a main focus of any preventive or therapeutic treatment. Accordingly, a sensorimotor training (SMT) approach should be preferred with regards to enhancing dynamic trunk stability and compensation capability (Hwang et al. 2013; Mannion et al. 2012; Searle et al. 2015; Saragiotto et al. 2016). More specifically, this indicates that people with a high risk of developing back pain should be prepared to compensate sudden, unexpected, repetitive loading of the trunk (Pedersen et al. 2004). After consideration of the above-presented results, the integration of perturbations in a sensorimotor trunk stabilizing intervention seems feasible (Mannion et al. 2012; Müller et al. 2016a; 2016b; Pedersen et al. 2004; Wirth et al. 2016). Besides, Milosevic et al. (2016) described significantly reduced response times (10-20 %) of the trunk muscles in response to expected perturbations. As a consequence, a sensorimotor perturbation training of the trunk muscles can contribute to an improvement of the response time as well as the ability to anticipate high strain situations. In addition, the selection of perturbations used should be applied by the limbs, but also directly to the torso, depending on which trunk muscles are being targeted. The use of functional and therefore highly dynamic exercises would thus be preferable. On the one hand, perturbations can be implemented through the use of unstable surfaces (Calatayud et al. 2015; Wolburg et al. 2015). On the other hand, the use of external, active stimuli could cause the targeted responses. In line with the results presented, not one particular type of perturbations is recommended for use with SMT treatment. This arises out of the different neuromuscular reaction patterns of the trunk demonstrated in response to different perturbations (magnitude and direction) (Müller et al. 2016). In addition, a regular change in the perturbations' characteristics is necessary, depending on the training aims, to account for the known quick adaptation mechanisms of the neuromuscular system. Furthermore, it should be noted for the application of lower limb perturbations that the (lateral ventral/dorsal) trunk muscles on the perturbed side make a decisive contribution to the active compensation of the balance (McGill et al. 2013; Mueller et al 2016). As a result, the point of application of the perturbation as well as the trunk muscles addressed during trunk-stabilising SMT should be positioned on the ipsilateral side of the body (Mueller et al 2016).

Another effect of back pain is an impairment of the trunk posture during gait that can be interpreted as a relieving posture. For therapy and prevention of BP, this means that, along with a SMT approach, also the trunk posture should be trained with and without application of external perturbations.

10.3 Limitations

For the interpretation and conclusion of the results presented above, some minor points remain to be considered and discussed. For the development of a measurement set-up to assess trunk stability, the study analysed middle-aged, normal physically active adults with and without back pain. Therefore, the validity of generalising the results to other cohorts is still unclear. Whether the loading situations developed are applicable to special cohorts to the same extent and with the same results, remains unsolved, e.g., athletes or older people.

For the analysis of the influence of back pain on trunk stability, it has to be mentioned that the group of patients did not include many high-pain patients with high chronification rates. The applicability of loading situations with unexpected, highly dynamic perturbations on such a population remains unclear. Nevertheless, altered neuromuscular and kinematic control of the trunk could be shown in the back pain patients, with Korff grades 2 to 4, enrolled in our study. Already in this cohort, it can be interpreted as reduced trunk stability, therefore leading to a greater risk of developing complaints or injury recurrence.

The suitability of the developed measurement set-up “trunk loading”, consisting of kinematic and electromyographic methods in continuous and sudden loading situations where the load is applied to the trunk by the limbs, could be proven from this thesis (Muellet et al. 2016; Müller et al. 2016a, 2016b). Transferability of the measurement methods and loading situations used in the clinical routine might be to complex. In addition, the costs for the equipment, the special demands for setting and staff, and for the standardization of data assessment, analysis and processing could hamper the transfer. Only the use in individual case analyses could be recommended. Moreover, application in the clinical routine assumes an adaptation of the set-up. For the measurement of the trunk motion, the suitability of very small wireless, portable measuring units based on inertial sensors has been proven (Bauer et al. 2015a; 2015b). For the assessment of the neuromuscular activity of the trunk muscles, a reduction of the 12-lead EMG set-up seems inevitable for clinical application. Nevertheless, in accordance with the presented results, both ventral and dorsal muscles should be integrated when analysing neuromuscular trunk muscle activity. A suitable loading situation that simultaneously accounts for the clinical setting remains to be developed, as well as highly dynamic loading situations that include perturbations.

Finally, it cannot be clarified in the current thesis whether the altered neuromuscular and kinematic parameters of trunk stability in BP patients are a result or a cause of back pain.

11. Conclusion: Clinical application and perspective

A reliable and valid measurement set-up for the assessment of the kinematic and neuromuscular response to continuous and sudden loading in healthy and back pain patients could be developed. This diagnostic approach consisted of loading situations in which sudden, external, unexpected perturbations were applied by the limbs (e.g., stumbling while walking). Neuromuscular activity (12-lead EMG) and 3D kinematics during the compensation of trunk loading were measured. Moreover, the need for an anatomical segmental kinematic model to analyse trunk motion was demonstrated. Consequently, the measurement set up to evaluate trunk stability in healthy individuals and back pain patients consisted of:

(1) Loading tasks:

- Highly dynamic, everyday tasks including sudden and unexpectedly applied perturbations generated by the limbs (e.g., walking with perturbation), and

(2) Measurement methods:

- Assessment of the neuromuscular response to perturbations (e.g., response times)
- Assessment of the kinematic response to perturbations (e.g., trunk posture).

Further, the influence of back pain on trunk function could be demonstrated in sudden but everyday like loading situations. A correlation between back pain and reduced trunk stability, quantified by the neuromuscular response and the trunk motion during the compensation of sudden, external loads could be shown. BP patients demonstrate a delayed muscular reaction as well as a (kinematic) relieving posture while compensating external perturbations during walking.

The transfer of the set-up in the clinical routine, however, requires further validation to allow transmission and adaptation of the complex methods including less complex loading situations. Besides this, the identification of people at high risk for back pain is of interest. The analyses of the underlying mechanisms of back pain on trunk loading and stability should be further emphasized. In addition, the magnitude of the trunk displacement and the compensation capacity of other body segments (leg, pelvis) should be analysed and calculated while stumbling.

For the prevention and treatment of back pain, sensorimotor training interventions to enhance stabilisation of the trunk muscles seem to be of great interest. The implementation of perturbations in sensorimotor-based interventions therefore appears to be feasible.

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Author´s contribution

The present thesis is designed as a publication-based dissertation. In this regard, four scientific articles have been submitted to peer-reviewed journals and currently under revision and/or published (accepted for publication). According to the local doctoral degree regulations (§ 7 (4), sentence No. 2), significant contributions to the articles from the respective co-authors were acknowledged and finally confirmed by each co-author:

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Study	**Design**	**Data Collection**	**Data analyses**	**Interpretation**	**Manuscript**
Paper 1 (Ch. 5.2)	**JM**, TE, FM	**JM**, TE, SK	**JM**, TE, SM	**JM**, HB, FM, SM	**JM**, HB, TE, FM, SM
Paper 2 (Ch. 6)	**JM**, HB, FM, SM	**JM**, JS	**JM**, JS	**JM**, HB, FM, SM	**JM**, HB, FM, SM, MR, JS
Paper 3 (Ch. 7)	**JM**, TE, FM, SM	**JM**, TE, AR	**JM**, TE, AR	**JM**, HB, SM, FM	**JM**, HB, TE, FM, SM, AR,
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Note: First author is highlighted in bold

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Abbreviations

3D	three-dimensional
BP	back pain
BPP	back pain patients
EMG	electromyography
EO	M. abdominis externus obliquus
G	(normal) gait
H	healthy controls
IO	M. abdominis internus obliquus
LA	lumbar area (segment)
LES	M. erector spinae L3
LD	M. latissimus dorsi
LTA	lower thoracic area (segment)
RA	M. rectus abdominis
ROM	range of motion
S	stumbling
SMT	sensorimotor training
UTA	upper thoracic area (segment)
UES	M. erector spinae T9

Acknowledgement

Firstly, I would like to express my thankfulness to my advisor Prof. Dr. Frank Mayer for providing me with the opportunity to become a scientist and moreover, to realize my PhD thesis at the University Outpatient Clinic Potsdam. I appreciate his leadership and encouragement to develop my scientific rational and to stay on my own two scientific legs. Huge thanks for making me a part of your team.

My first steps in the scientific world were accompanied by the outstanding personalities Anja, Carlsohn, Heiner Baur, Friederike Scharhag-Rosenberger, Jürgen Scharhag and Kathrin Steffen. I really appreciate learning from all of you! Thank you for your guidance. You all influenced my personal way of scientific thinking. Thank you! I am looking forward to our future scientific and personal exchanges.

Further, I would like to thank all colleaques at the University Outpatient Clinic for the teamwork and the inspiring atmosphere during my work as a research assistant. A personal and special thank you to Christoph Otto, Michael Cassel, Peggy Kotsch and Katja Fröhlich. Josi, thanks for being such a good fellow during the last years and always chearing me up in wild but not always evidence-based discussions ;-). I really appreciate the time with you and I will never stop dreaming of our “Pediatric Clinic” at the Baltic Sea ;-). Tilman, thanks for letting us do this big study together. It was an apleasure for me measuring, analysing and discussing the bundge of data with you! I guess we rocked and should let the whole world stumble in the future ;-). Anika and Ina, thanks for taking care of all the administrative stuff from the beginning. You eased up my work and allowed the focus on the real things.

My deepest appreciation goes to all co-authors of my original studies for their partnership, hard work and scientific passion. Martin Löhrer, thank you so much for guiding me through my kinematic data and always answering my bunch of questions. In addition, I express my thankfulness to Antje Reschke, Stephan Kopinksi, Martin Wolter, Gerrit Hain, Monique Wochatz and Konstantina Intziegianni for their assistance with data assessment and analysis. Hannes Kaplick, you really made my kinematics shine in beautiful curves, thanks for this. Furthermore, all participants donated to the successful achievements of my research project. Thank you all so much for your valuable time!

My warmest and special grateness goes to my entire family. Words cannot express how grateful I am, especially to my mother, Birgit, and father, Wolfgang, for all your strength, believes and guidance during my whole life. I love you! Thanks for showing me your entire desire for sports in all it´s kinds

and therefore, laying the fundamental foundation of my scientific desire. Grandpa, thank you so much for teaching me your passion and I will never forget: life is now a tree at the sea.
I would also like to thank all of my friends who supported me and strengthened my back besides the path of the scientific life. Nancy, Dörte, Antje, Janine, I am so happy to have you by my side.

My highest gratitude goes to the two most important people in my life: Steffen and Rafael! Steffen, thanks for taking me into your world, guiding me and reaching me your hands while taking my first scientific steps. You always strengthen my back and providing me a heaven of peace in this incredibly fast running world. We are a perfect team, 100% evidence-based ☺. Rafa, you lighten up our world and make me the proudest of all Mums. Thank you so much for teaching us that life is more than just working in the lab! I love you both!

Affidavits according to doctoral degree regulations (§ 4 (2), sentences No. 4 and 7) of the Faculty of Human Sciences, University of Potsdam:

Hereby, I declare that this thesis entitled *“Trunk loading and Back Pain - Three-dimensional motion analysis and evaluation of neuromuscular reflex activity in response to sudden and continuous loading ”* or parts of the thesis have not yet been submitted for a doctoral degree to this or any other institution neither in identical nor in similar form. The work presented in this thesis is the original work of the author. I did not receive any help or support from commercial consultants. All parts or single sentences, which have been taken analogously or literally from other sources, are identified as citations. Additionally, significant contributions from co-authors to the articles of this cumulative dissertation are acknowledged in the authors’ contribution section.

______________________________ ______________________________

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