ONSETS AND DEPENDENCIES
OF STRENUOUS SPINE BENDING ACCELERATIONS
IN DROP LANDINGS

AUFTRETEN UND ABHÄNGIGKEITEN VON
BELASTENDEN WINKELBESCHLEUNIGUNGEN AN DER WIRBELSÄULE
BEI SPRUNGLANDUNGEN

BY

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AFFIDAVITS

According to the 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 "**ONSETS AND DEPENDENCIES OF STRENUOUS SPINE BENDING ACCELERATIONS IN DROP LANDINGS**", 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.

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ABSTRACT

BACKGROUND: Physical activity involving high spinal load has been exposed to possess a crucial impact in the genesis of acute and chronic low back pain and disorder. Vigorous spinal loads are surmised in drop landings, for which strenuous bending loads were formerly evinced for the lower extremity structures. Thus far, clinical studies revealed that repetitive landing impacts can evoke benign structural adaptions or damage to the lumbar vertebrae. Though, causes for these observations are hitherto not conclusively evinced; since actual spinal load has to date not been experimentally documented. Moreover, it is yet undetermined how physiological activation of trunk musculature compensates for landing impact induced spinal loads, and to which extend trunk activity and spinal load are affected by landing demands and performer characteristics. **AIMS** of this study are 1. the localisation and quantification of spinal bending loads under various landing demands and 2. the identification of compensatory trunk muscular activity pattern, which potentially alleviate spinal load magnitudes. Three consecutive Hypotheses (***H1 - H3***) were hereto postulated: ***H1*** posits that spinal bending loads in segregated motion planes can feasibly and reliably be evaluated from peak spine segmental angular accelerations. ***H2*** furthermore assumes that vertical drop landings elicit highest spine bending load in sagittal flexion of the lumbar spine. Based on these verifications, a second study shall prove the successive hypothesis (***H3***) that diversified landing conditions, like performer's landing familiarity and gender, as an implementation of an instantaneous follow-up task, affect the emerging lumbar spinal bending load. Herein it is moreover surmised that lumbar spinal bending loads under distinct landing conditions are predominantly modulated by herewith disparately deployed conditioned pre-activations of trunk muscles. **METHODS:** To test the above arrayed hypothesis, two successive studies were carried out. In **STUDY 1**, 17 subjects were repetitively assessed performing various drop landings (heigth: 15, 30, 45, 60cm; unilateral, bilateral, blindfolded, catching a ball) in a test-retest-design. Herein individual peak angular accelerations [α_{MAX}] were derived from three-dimensional motion data of four trunk-segments (upper thoracic, lower thoracic, lumbar, pelvis). α_{MAX} was herein assessed in flexion, lateral flexion, and rotation of each spinal joint, formed by two adjacent segments. Reliability of α_{MAX} within and between test-days was evaluated by *CV%, ICC 2.1, TRV%,* and Bland & Altman Analysis (*BIAS±LoA*). Subsequently, peak flexion acceleration of the lumbo-pelvic joint [$\alpha_{FLEX[LS-PV]}$] was statistically compared to α_{MAX} expressions of each other assessed spinal joint and motion plane (*Mean ±SD, Independent Samples T-test*). **STUDY 2**

deliberately assessed mere peak lumbo-pelvic flexion accelerations [$\alpha_{FLEX[LS\text{-}PV]}$] and electro-myographic trunk pre-activity prior to $\alpha_{FLEX[LS\text{-}PV]}$ on 43 subjects performing varied landing tasks (height 45cm; with definite or indefinite predictability of a subsequent instant follow up jump). Subjects were contrasted with respect to their previous landing familiarity (>1000 vs. <100 landings performed in the past 10 years) and gender. Differences of $\alpha_{FLEX[LS\text{-}PV]}$ and muscular pre-activity between contrasted subject groups as between landing tasks were equally statistically tested by three-way mixed *ANOVA* with *Post-hoc tests*. Associations between $\alpha_{FLEX[LS\text{-}PV]}$ and muscular pre-activity were factor-specifically assessed by Spearman's rank order correlation coefficient (r_S). Complementarily, muscular pre-activity was subdivided by landing phases [*DROP, IMPACT*] and discretely assessed for phase specific associations to $\alpha_{FLEX[LS\text{-}PV].}$ Each muscular activity was moreover pairwise compared between *DROP* and *IMPACT* (*Mean ±SD, Dependent Samples T-test*). **RESULTS:** α_{MAX} was presented with overall high variability within test-days (*CV =36%*). Lowest intra-individual variability and highest reproducibility of α_{MAX} between test-days was shown in flexion of the spine. $\alpha_{FLEX[LS\text{-}PV]}$ showed largely consistent sig. higher magnitudes compared to α_{MAX} presented in more cranial spinal joints and other motion planes. $\alpha_{FLEX[LS\text{-}PV]}$ moreover gradually increased with escalations in landing heights. Landing unfamiliar subjects presented sig. higher $\alpha_{FLEX[LS\text{-}PV]}$ in contrast to landing familiar ones (*p=.016*). M. Obliquus Int. with M. Transversus Abd. (*66 ±32%MVC*) and M. Erector Spinae (*47 ±15%MVC*) presented maredly highest activity in contrast to lowest activity of M. Rectus Abd. (*10 ±4%MVC*). Landing unfamiliar subjects showed compared to landing familiar ones sig. higher activity of M. Obliquus Ext. (17 *±8%MVC, 12 ±7%MVC, p= .044*). M. Obliquus Ext. and its co-contraction ratio with M. Erector Spinae moreover exhibited low but sig. positive correlations to $\alpha_{FLEX[LS\text{-}PV]}$ (r_s*=.39,* r_s*=.31*). Each trunk muscule distributed larger shares of its activity to *DROP*, whereas peak activations of most muscles emerged in the proportionally shorter *IMPACT* phase. Commonly increased muscular pre-activation particularly at *IMPACT* was found in landings with a contrived follow up jump and in female subjects, whereby $\alpha_{FLEX[LS\text{-}PV]}$ was hereof only marginally affected. **DISCUSSION:** Highest spine segmental angular accelerations in drop landings emerge in sagittal flexion of the lumbar spine. The compensatory stabilisation of the spine appears to be preponderantly provided by a dorso-ventral co-contraction of M. Obliquus Int., M. Transversus Abd. and M. Erector Spinae. Elevated pre-activity of M. Obliquuis Ext. supposably characterises poor landing experience, which might engender increased bending loads to the lumbar spine. A pervasive large variability of spinal angular accelerations measured across all landing types, suggests a

multifarious utilisation of diverse mechanisms compensating for spinal impacts in landing performances. A standardised assessment and valid evaluation of landing evoked lumbar bending loads is hereof largley confined. **CONCLUSION**: Drop landings elicit most strenuous lumbo-pelvic flexion accelerations, which can be appraised as representatives for high energetic bending loads to the spine. Such entail the highest risk to overload the spinal tissue, when landing demands exceed the individual's landing skill. Previous landing experience and training appears to effectively improve muscular spine stabilisation pattern, diminishing spinal bending loads.

ZUSAMMENFASSUNG

HINTERGRUND: Wirbelsäulenbelastungen in Alltagssituationen und während sportlicher Belastung kommt eine hohe Bedeutung mit Blick auf die Entstehung und das Weiterbestehen von akuten und chronischen Rückenbeschwerden zu. Kritisch hohe Wirbelsäulenbelastungen werden bei Sprunglandungen angenommen, während hierzu hochintensive exzentrische Belastungen bislang lediglich für anatomische Strukturen der unteren Extremität nachgewiesen wurden. Vorangegangene klinische Studien konnten zeigen, dass repetitive Landungsstöße sowohl eine strukturelle Anpassung, als auch morphologische Schäden der Lendenwirbelkörper hervorrufen können. Valide Ursachen für diese Beobachtungen sind bislang wissenschaftich nicht abschließend belegt, insbesondere da der experimentelle Nachweis für die hierin vermuteten tatsächlichen Wirbelsäulenbelastungen fehlt. Darüber hinaus ist nicht geklärt in wieweit die physiologisch kompensatorische Aktivierung der Rumpfmuskuatur Einfluß auf die Wirbelsäulenbelastung bei Landungen nehmen, und wie stark Landungs- und Personencharakteristika die Rumpfaktivierung und Lendenwirbelsäulenbeastungen beeinflussen. **ZIELSETZUNGEN:** Ziele der Untersuchungen sind 1. die Lokalisierung und Quantifizierung von Biegebelastungen der Wirbelsäule unter verschiedenen Landungsbedingungen und 2. die Identifizierung muskulärer Kompensationsmechanismen des Rumpfes, welche das Belastungsausmaß an der Wirbelsäule möglicherweise modulieren. Hierzu wurden drei Hypothesen (***H1* – *H3***) formuliert. In ***H1*** wird postuliert, dass Biegebelastungen in einzelnen Bewegungsebenen der Wirbelsäule als maximale Winkelbeschleunigungen zwischen Wirbelsegmenten, auf der Basis kinematischer Daten, valide und reliabel abgeleitet und evaluiert werden können. In ***H2*** wird angenommen, dass bei vertikalen Sprunglandungen die höchsten Wirbelsäulenbelastungen in der sagittalen Beugung der Lendenwirbelsäule auftreten. Aufbauend auf den Ergebnissen dieser Hypothesen soll eine Folgestudie die Annahme (***H3***) belegen, dass Landungsbedingungen, wie Vorerfahrungen mit Sprunglandungen, Geschlecht, sowie die Absicht zu unmittelbaren Anschlussbewegungen, die auftretenden Lendenwirbelsäulenbelastungen beeinflussen. Hierzu wird postuliert, dass auftretende Biegebelastungen, in Abhängigkeit obiger Landungsbedingungen, auf einen unterschiedlichen Einsatz von vorwiegend konditionierten muskulären Kompensationsmechanismen des Rumpfes zurückzuführen sind. **METHODE**: Zur Überprüfung der Hypothesen wurden zwei sukzessive Studien durchgeführt. In **STUDIE 1** wurden, zur Repräsentation von Wirbelsäulenbiegebelastungen, 17 Probanden wiederholt bei verschiedenen Landungen (15, 30, 45, 60cm Höhe; einbeinig, beidbeinig, verblindet,

beim Fangen eines Balles) in einem Test-Retest-Design gemessen. Hierin wurden individuelle maximale Winkelbeschleunigungen [α_{MAX}] aus drei-dimensionalen Bewegungsdaten zwischen insgesamt 4 Rumpfsegmenten (oberes thorakales-, unteres thorakales-, Lendenwirbelsäulen-, und Becken-Segment) abgeleitet. α_{MAX} wurde hierbei jeweils im Gelenk zwischen zwei benachbarten Segmenten in Flexion, Lateralflexion und Rotation erfasst. Die Reliabilität von α_{MAX} innerhalb und zwischen den Messtagen wurde mittels *CV%, ICC 2.1, TRV%,* und *Bland & Altman Analyse* (*BIAS±LoA*) quantifiziert. In Folge wurden α_{MAX} zwischen dem lumbalen- und dem Beckensegment in der Flexion [$\alpha_{FLEX[LS\text{-}PV]}$] mit allen weiteren gemessenen Segmenten und Bewegungsebenen gegenübergestellt (*Mean ±SD, T-Test* für unabhängige Stichproben). In **STUDIE 2** wurden geziehlt zuvor eruierte höchste maximale sagittale Beugungsbeschleunigung der Lendenwirbelsäule [$\alpha_{FLEX[LS\text{-}PV]}$] und elektro-myografische Rumpfaktivität vor dem Auftreten von $\alpha_{FLEX[LS\text{-}PV]}$ während unterschiedlicher Landungen (Höhe 45cm; mit und ohne planbaren Anschlusssprung) an 43 Probanden erfasst. Die Probanden unterschieden sich bezüglich ihrer Landungsvorerfahrung (>1000 vs. <100 Landungen in den letzten 10 Jahren) und ihres Geschlechtes. Unterschiede zwischen Landungsvorerfahrung und Geschlecht sowie zwischen unterschiedlichen Landungstypen wurden gleichermaßen durch dreifaktorielle *ANOVA* mit *Post-hoc Tests* für $\alpha_{FLEX[LS\text{-}PV]}$, und muskuläre Voraktivierung getestet. Abhängigkeiten von $\alpha_{FLEX[LS\text{-}PV]}$ zu muskulärer Voraktivierung des Rumpfes wurde durch faktorspezifische Rangkorrelationsanalyse (r_s) berechnet. In der Folge wurden muskuläre Rumpfaktivitäten in Landephasen [*DROP, IMPACT*] unterteilt und analog im Einzelnen nach ihren Assoziationen zu $\alpha_{FLEX[LS\text{-}PV]}$ getestet. Zudem wurde jegliche Muskelaktivierung paarig zwischen *DROP* und *IMPACT* verglichen (*Mean ±SD, T-Test* für abhängige Stichproben). **ERGEBNISSE:** Die Ausprägung von α_{MAX} zeigte insgesamt hohe Variabilität innerhalb eines Testtages (*CV =36%*). Geringste intra-individuelle Variabilität und zugleich höchste Reproduzierbarkeit zwischen den Test-Tagen wurde für α_{MAX} in Flexion der Wirbelsäule gefunden. $\alpha_{FLEX[LS\text{-}PV]}$ zeigte nahezu durchgehend sig. höhere Werte im Vergleich zu α_{MAX} kranialerer Gelenke und anderer Bewegungsebenen. $\alpha_{FLEX[LS\text{-}PV]}$ stieg zudem graduell mit zunehmenden Landungshöhen. Landungsunerfahrene Probanden wiesen im Vergleich zu Probanden mit Vorerfahrung signifikant höhere $\alpha_{FLEX[LS\text{-}PV]}$ auf (*p=.016*). Markant höchste muskuläre Aktivität wurde von M. Obliquus Int. mit M. Transversus Abd. (*66 ±32%MVC*) und M. Erector Spinae (*47 ±15%MVC*), verglichen zu geringster Aktivität von M. Rectus Abd. (*10 ±4%MVC*) dargeboten. Bei Landungsunerfahrenen wurde im Vergleich zu Landungserfahrenen eine sig. höhere Aktivität des M. Obliquus Ext. gemessen (17 *±8%MVC, 12 ±7%MVC, p= .044*).

Zudem konnten schwache aber sig. positive Korrelation zwsichen der Aktivität des M. Obliquus Ext. bzw. dessen Kokontraktion mit dem M. Erector Spinae zu $\alpha_{FLEX[LS\text{-}PV]}$ nachgewiesen werden (r_s=.39, r_s=.31). Die Rumpfmuskulatur zeigte insgesamt anteilig mehr Bereitstellung während *DROP*, wobei Spitzenaktivitäten nahezu aller Rumpfmuskeln in der proportional kürzeren *IMPACT*-Phase auftraten. Frauen und Landungen mit geplantem unmittelbarem Anschlussspung zeigten insgesamt höhere Voraktivierung der Rumpf-muskulatur, vorallem in *IMPACT*, wobei sich $\alpha_{FLEX[LS\text{-}PV]}$ unter diesen Bedingungen nur insignifikant von anderen Landungen unterschied. **DISKUSSION:** Bei Landungen treten höchste segmentale Winkelbeschleunigungen in sagittaler Beugung der Lendenwirbelsäule auf. Die kompensatorische Stabilisation des Rumpfes scheint dabei maßgeblich durch eine dorso-ventrale Kokontraktion des M. Obliquus Int., M. Transversus Abd. und dem M. Erector Spinae zu erfolgen. Eine hohe Voraktivierung des M. Obliquuis Ext. kann als Maß einer geringen Landungserfahrung diskutiert werden und führt möglicherweise zu erhöhten Biegebelastungen an der Lendenwirbelsäule. Die in allen untersuchten Landungen dargebotene hohe Variabilität gemessener Winkelbeschleunigungen lassen auf sehr variabel eingesetzte Impulskompensationsmechanismen bei der Durchführungen von Landungen schließen. Eine standardiserte Erfassung und valide Einschätzung von Biegebelastungen der Lendenwirbelsäule bei Sprunglandungen ist hierdurch stark eingeschränkt. **SCHLUSSFOLGERUNG:** Sprunglandungen verursachen höchst belastende segmentale Winkelbeschleunigungen an der Lendenwirbelsäule, vorrangig in der Flexion. Diese können physiologisch bedingt als Maß für hoch energetische Biegebelastungen der Wirbelsäule verstanden werden. Ein mögliches Risiko hieraus resultierender struktureller Überlastung muss insbesondere in Betracht gezogen werden, wenn Landungsanforderungen die individuellen Landungfähigkeiten übersteigen. Eine probate muskuläre Wirbelsäulen-stabilisation bzw. derer regelmäßiges Training in der Durchführung von Landungsvorgängen scheint erforderlich um auftretende Biegebelastungen zu reduzieren.

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[α_{MAX}]	peak segmental angular acceleration
[$\alpha_{FLEX[LS\text{-}PV]}$]	peak lumbo-pelvic flexion acceleration
[ANOVA]	analysis of variances
[ASI]	anterior spina iliaca
[BF]	blindfolded landing
[Bias]	systematic error = mean of the differences
[BL]	bilateral landing
[BMI]	body mass index
[CCI]	co-contraction index
[CMC]	coefficient of multiple correlations
[COM]	centre of mass
[COM_{MIN}]	minimal vertical centre of mass position
[CV]	Coefficient of Variation (intra-individual)
[CV_{AV}]	Coefficient of Variation (inter-individual)
[Diff.]	absolute difference
[DJ]	drop jump
[DL]	drop landing
[DS]	delayed choice drop landing / drop jump
[EMG]	electromyography
[EO]	musculus Externus Abdominis
[ES]	musculus Erector Spinae
[FLEX]	sagittal segmental flexion
[GRF]	ground reaction force
[GRF_{BW}]	peak vertical ground reaction force as ratio to subject's bodyweight
[GRF_{N}]	peak vertical absolute ground reaction force
[ICC]	intra-class correlation coefficients
[IGC]	initial ground contact
[IO]	musculus Internus Abdominis
[LAT]	coronal segmental lateral bending
[LoA]	limits of agreement
[LS]	lumbar spine segment
[LS-PV]	lumbar spine to pelvis joint
[LTS]	lower thoracic spine segment
[LTS-LS]	lower thoracic to lumbar spine joint
[MAX]	maxima
[MIN]	minima
[MVC]	maximal voluntary contraction
[OH]	overhead catch landing
[PRE]	landing period from peak rise height to peak lumbo-pelvic flexion acceleration
[PSI]	posterior spina iliaca

[PV]	pelvis
[PV-GF]	global angulation of pelvis to ground floor
[RA]	musculus Rectus Abdominis
[RMS]	root mean square
[ROM]	range of motion
[ROT]	transverse segmental rotation
[r]	Pearson's correlation coefficient
$[r_S]$	Spearmans's rank order correlation coefficient
[sEMG]	surface electromyography
[t-IGC]	time of initial ground contact
[TrA]	musculus Transversus Abdominis
[TRV]	test-retest variability
$[t\text{-}\alpha_{FLEX[LS\text{-}PV]}]$	temporal occurrence of peak lumbo-pelvic flexion acceleration
$[t\text{-}\alpha_{MAX}]$	time of peak segmental angular acceleration
$[t\text{-}IGC<t\text{-}\alpha_{MAX}]$	occurrence of peak segmental angular acceleration after onset of initial ground contact
$[t\text{-}IGC<t\text{-}\alpha_{FLEX[LS\text{-}PV]}]$	occurrence of peak lumbo-pelvic flexion acceleration after onset of initial ground contact
[UL]	unilateral landing
[UTS]	upper thoracic spine segment
[UTS-LTS]	upper thoracic to lower thoracic spine joint
[%Diff.]	relative difference

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INTRODUCTION

I - CLINICAL IMPORTANCE AND PREVALENCE OF SPINAL LOAD IN LANDINGS

Mechanical load to the spine has become a versatilely investigated aspect to the aetiology of spinal traumatic injury and low back pain, (*Garg & Moore, 1992; Marras et al., 1993; Norman et al., 1998*). Whilst causes of spinal injuries are most frequently referred to high impact trauma, most onsets of low back pain are believed to result from degeneration of spinal soft tissue due to preceding singular or frequent overload (*Stokes & Iatridis, 2004; Lawrence et al., 2006*) and dysponesis (*Whatmore & Kohi, 1968*). Overload of spinal tissue most frequently occurs at physically strenuous workplaces (*Garg & Moore, 1992; Marras et al., 1993; Yassi & Lockhart, 2013; Goswami et al., 2016*) and in exertions of high-load-physical activity and -sport (*Lawrence et al., 2006; Trompeter et al., 2017*), wherefore high rates of spinal pathologies are evinced (*Boden & Jarvis, 2008; Borg-Stein et al., 2012; Hume et al., 2013; de Jonge & Kramer, 2014*). Whilst spine overstraining loading mechanisms in occupational environments, which potentially lead to low back disorder and pain, have been identified in: the handling of heavy external weights, forced unnatural postures, and spine bending and twisting at either high frequencies or at high velocities and accelerations (*Garg & Moore, 1992; Marras et al. 1993; Yamamoto, 1997*); actual mechanisms causing low back pain in sport exerting populations are not as thoroughly exposed. However, high compressive impact load in sports has been shown to potentially harm the spines structural posture properties and its soft tissue components (*Stinson, 1993; Lawrence et al., 2006; Kruse & Lemmen, 2009; Daniels et al., 2011; Hume et al., 2013; Schmidt et al., 2013; de Jonge & Kramer, 2014; Mortazavi et al., 2015*). The of high impact overload emerging most prevalent spinal injuries are: vertebral fractures, intervertebral disc degenerations and herniations, and spinal ligamentous and muscular sprains, strains, and tears; each constituting to back pain (*Alexander, 1985; Keene & Drummond, 1985; Lawrence et al., 2006; Khan et al. 2008; Schmidt et al., 2013; Massel & Singh, 2017*). Severest impacts leading to such injuries often result from contact sports, such as football, ice hockey, wrestling, etc. (*Boden & Jarvis, 2008*), whilst non-contact sports feature highest spinal impacts in falls and landings (*Recknagel & Witte, 1996; Teh et al., 2003; Hume et al., 2013*).

In landings, spinal load and injury can occur as the spine is designated to transmit impact forces (*Alexander, 1985*), which are known to emerge on high magnitudes at the lower limb in the instant of ground contact (*Zhang et al., 2008*). Unfortunately, the presence of actual mechanical loads to the spinal structures in landings has hitherto only received sparse attention in research (*Recknagel & Witte. 1996; Ng et al., 2006; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*), consigning inconclusive outcomes. Though, several types of foot first landings have been supposed to generate greatly vigorous impacts, involving injury risks to the spinal structures (*Lowenstein et al., 1989; Recknagel & Witte, 1996; Teh et al., 2003; Hume et al., 2013*) or cause overuse induced spinal deformations (*Hume et al., 2013*). *Hume et al.* (*2013*) hereto postulated that common spinal injuries in gymnasts are likely to be evoked by high forces and decelerations, whereas peak forces of herein performed landings have been documented to reach up to an *11*-fold of gravitational force [*g*], emerging from extensive amounts of previously accumulated energy. *Recknagel & Witte* (*1996*) by contrast conducted a concealed erect-standing vertical drop experiment from lowest heights of *3* to *5cm*, whereof severe dorso-cranial accelerations at the lumbo-sacral region were calculated, creating shearing impact forces of up to *8g*. Such great accelerations and concomitant shear forces acting on the lumbar spine have in this context been supposed to promote spondylolisthesis (*Recknagel & Witte, 1996*). In a previous evaluation of spinal shear load, *Gallagher & Marras* (*2012*) reviewed limits lumbar spinal tolerances against shear forces. They herein concluded that: in order to warrant spinal safety, shear stress limits should be held below *1000N* for occasional exposure, while with regards to fatigue effects, repetitive exposure should be curtailed to a limit of *700N* shear when load exposure was more frequent than *100times/day*. The herein constituted tolerance retreat prompts further concerns in the appraisal of potential landing induced spinal overload, as landings in physical activity most commonly occur at high prevalence per exercise session. *Panther & Bradshaw* (*2013*) more recently assessed linear trunk accelerations and body masses of *50* male professional football players during a preseason training session. They hereof presented average peak vertical trunk compression forces of ≈*7g* evoked from landings of game-like maximal vertical jumps. Though, *Panther & Bradshaw* (*2013*) argued that their employed accelerometry might not be sufficiently accurate for an injury risk appraisal; however, their presented acceleration measures occurred on not far lower magnitudes than those previously concerned in context of severe spinal injuries (*Recknagel & Witte, 1996; Hume et al., 2013*). Notably, the exclusively determined sole linear accelerations in foregone landing research (*Recknagel & Witte. 1996; Ng et al., 2006; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*) do not provide a complete disclosure of potential loads and hazards to the spine. Instead, large versatility of landing induced spinal load manifestations is commonly conjectured, whereat compression, shear stress, and bending strain are surmised (*Recknagel & Witte, 1996; Kohrt et al., 2004; Santello, 2005*).

A versatile presence of high spinal loads induced by landing can be moreover inferred from several osteoporosis research findings. Herein, landings have been exposed and applied to create beneficial bone formation adaptions to the lower extremity, the pelvis and the lumbar spine (*Bassey et al., 1998; Heinonen et al., 2000; Fuchs et al., 2001; MacKelvie et al., 2002, 2003; Arnett & Lutz, 2002; Johannsen et al., 2003; Uusi-Rasi et al., 2003; Shibata et al., 2003; Kontulainen et al., 2002a, 2004b; McKay et al., 2005; Kato et al., 2006; Gunter et al., 2008; Weeks et al., 2008; Weeks & Beck, 2012; Guadalupe-Grau et al., 2009; Niu et al., 2010; Meyer et al., 2011; Winters-Stone et al., 2011; Bolton et al., 2012; Greenway et al., 2015; Maggio et al., 2012; Saarto et al., 2012; Sindel et al., 2014; Bolam et al., 2013, 2015; Hinton et al., 2015*). In this huge body of research, beneficial bone mineral content and -density gains have been evoked by jump- and landing interventions (*9* weeks to *18* month), wherein a large amount of jumps or sole landings (n ≈*3,700* to ≈*31,000*) was ascribed to various populations with osteopenia or osteoporosis. One other study showed similar vertebral bone gains elicited in pre-pubertal girls after 1 year of playing handball (*Vicente-Rodriguez et al., 2004*). However, few of the above listed studies have failed to show lumbar vertebral bone adaptions in discrepancy to observed lower extremity bone gains (*Arnett & Lutz, 2002; Johannsen et al., 2003; McKay et al., 2005*). Moreover, few other landing interventions didn't elicit any bone adaption (*Saarto et al., 2012; Bolam et al., 2013, 2015*). All in all, these findings are not sufficiently consistent to reveal if any particular characteristic within these interventions produced most recurrent vertebral bone adaptions. Nevertheless, the American College of Sports Medicine Position Stand on: "Physical activity and bone health" endorses high impact activities as gymnastics, jumping and plyometric exercises among others, in order to promote bone formation (*Kohrt et al., 2004*); even though a conclusive statement is missing about if bone formation effects result from either gravitational or muscular forces (*Kohrt et al., 2009*).

Subsumingly previous literature provides various scattered indications for spinal load concerns in the context of landing. However, these indications are most fragmentary in regards to actual load features, whereof yet no coherent conception of spinal load components and magnitudes with particular regard to injury hazards was yielded. The collectively scarce body of literature contextualising landings with spinal injury hazards might be predominantly ascribed to the prevailing lack of reliable load assessment methods. On the downside, considerably large prevalence of impactful landing executions are surmised in several types of professional sports, physical activity, and performance training (*Harreby et al., 1997; Louw & Grimmer, 2006; Vignon et al., 2006; Lesinski et al., 2014*). Thence, both, recreational and professional jumpers are thereby frequently challenged for essential landing performances and therewith entailed load to the spine.

II - Physiological Control of Landing Impacts

Executions of foot first landings are known to require multi-segmental feed-forward neuro-motor modulations of joint positions and muscle-tendon stiffness to prepare dissipation and absorption of the impact with touch down. In immediate succession, additional sensory-motor activity is elicited to counteract the perceived eccentric joint rotating impact magnitudes, in order to decelerate all affected body segments and stabilise body posture (*Santello, 2005*), (*Figure I*). Whilst the actual integral interaction between predictive and reactive components of muscle activity modulation in landings is yet not fully understood, few physiological concepts of muscular activity behaviour have been exposed by previous research.

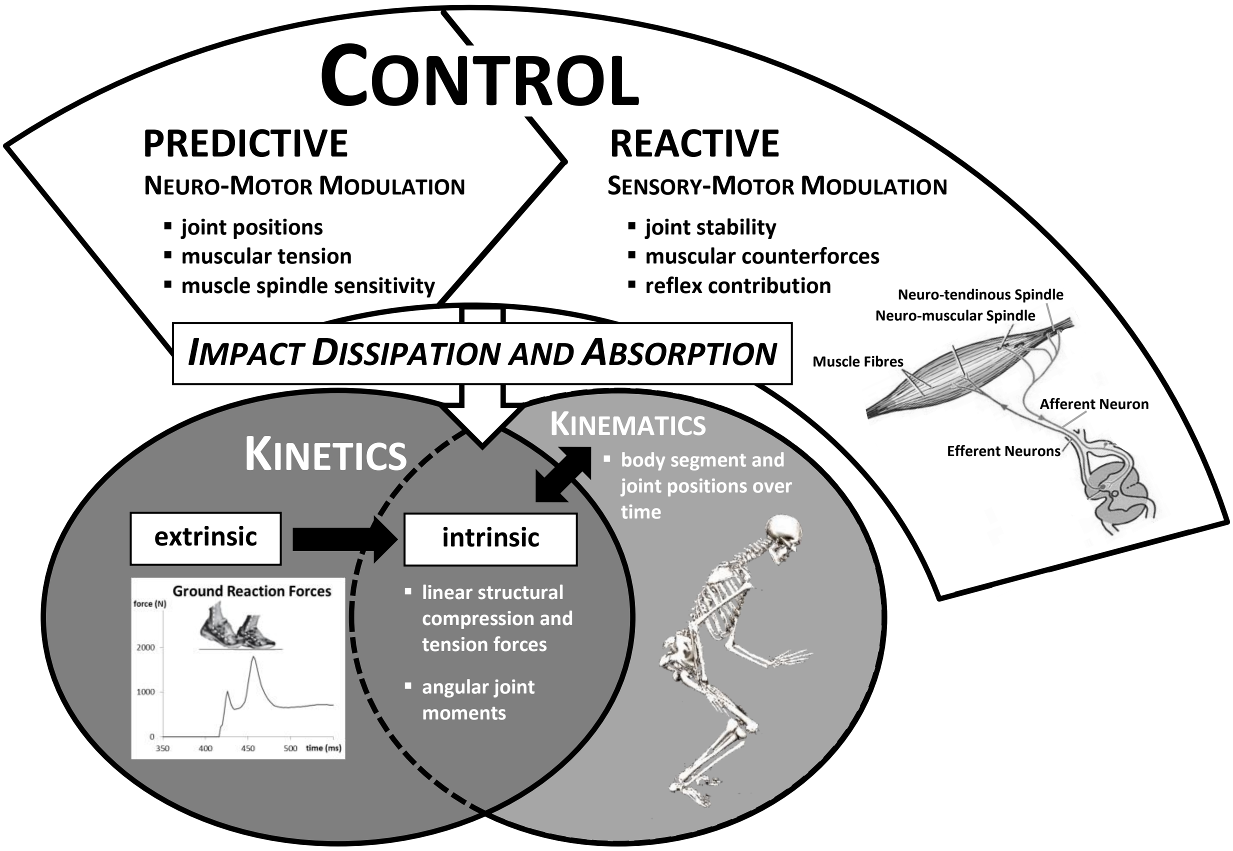


Figure I: Physiological control of landing impacts

Preparatory Control.

Essential preparatory joint stiffening lower limb activation of muscles has been shown to build up by around *120* to *100ms* prior to ground contact. The magnitude of muscle activity evoked in this time frame appears to be approximately linearly scaled to drop height (*Santello, 2005*). The configuration of preparatory muscular activity volume herein is implemented by recruiting a desired quantity of motor units or by adjusting neuronal firing frequencies. The accurate compliance of preparatory muscle activity magnitudes with the landing demands is believed to be initially depending on visual information or acquired sensory-motor memories, if applicable (*Santello, 2005*). Deployment of preparatory muscle activations on both, lower limb joint flexor and extensor muscles during landing has been well documented (*Pflum et al., 2004; Kulas et al., 2010; Iida et al., 2011, 2012; Prieske et al., 2013, 2015*), whist inconsistency prevails about a holistic posture stabilisation incorporating trunk muscle activity (*Santello, 2005; Kulas et al., 2006; Iida et al., 2011; Prieske et al., 2015b; Haddas et al., 2015, 2016a, b*). Within this literature dissent, *Santello* (*2005*) initially implied that preparatory muscle activity is integrally expected in any movement where impacts with external objects are expected. With regard to the previously acknowledged impact transmission role of the trunk (*Alexander, 1985*), it is consequently suggested that trunk muscle pre-activity should attend the preparation of any impact, prefaced by the lower limbs at the landing's touch down. However this supposition was not consistently affirmed by later investigations. Hereto *Iida et al.* (*2011*) presented an apparent reduction of abdominal muscle activity within the final ≈*50ms* prior to initial ground contact, along with a rather reactive response of Erector Spinae in succession to touch down. *Prieske et al.* (*2013*) instead showed quite equal levels of Rectus Abdominis and Erector Spinae activity in the drop-phase of landing. Besides, *Kulas et al.* (*2006*) supposed gender specific abdominal pre-activation strategies with males favouring a feed-forward stabilisation strategy, which was not shown by females. Upon several discordances between these studies yet investigating involuntary preparatory trunk activity in unrestraint drop landing (*Kulas et al., 2006; Iida et al., 2011; Prieske et al., 2013*), the trunk's true preparatory conduct in landings is to date unclear. Though, a general but condition specific implementation of trunk muscular pre-activation for landing control can be surmised, which compares with previous findings for lower limb muscular pre-activation (*Devita & Skelly, 1992; Blackburn et al., 2009; Iida et al., 2011; Iida et al., 2012; Prieske et al., 2013*). Unlike to findings on muscular pre-activity, kinematic landing observations have indicated landing height- and gender-independent pre-configurations of lower limb joint angles to rather default extended positions (*Devita & Skelly, 1992; Zhang et al., 2000; Santello, 2005*). This apparent standardised pre-modulation is believed to provide larger exploitable range of motion to the implementation of impact damping in succession to touch down (*Devita & Skelly, 1992; Santello, 2005*). Hereto scientific literature does not provide any valuation of the trunk's landing kinematics.

Reactive Control.

In succession of touch down, each lower limb joint starts to flex, absorbing and dissipating the landing's impact energy by eccentric muscular work (*Devita & Skelly, 1992; Zhang et al., 2000; Blackburn & Padua, 2008*). Herein the exploited joint motion ranges are generally in a positive relationship to the overall kinetic energy of the landing (*Zhang et al., 2000*). Actual mechanisms of muscular post landing impact control are not fully unveiled (*Santello, 2005*); though a common contemplation prevails that impact energy dissipation and absorption at the lower limb joints and muscles are achieved by the synergy of conditioned volitional and reflexive muscular activity (*Santello, 2005*). Herein previously established musculo-tendinous stiffness is utilised, allowing a proportionate absorption of impact energy by transforming it into tissue stretch, which is modulated by eccentric muscular work. The additionally introduced short and long latency reflexes, which are evoked by neuro-muscular and neuro-tendinous spindle stretches, are believed to support the damping process with supplementary reactive fibre tensioning, thus increasing muscular work (*Santello, 2005*). Hereto it is believed that the herein dominant share in early impact absorption is ascribed to conditioned reflexes, which occur in a pre-defined time course. This supposition is physiologically substantiated, since latencies of spindle stretch reflexes have been found with delays, longer than the short time course at which peak landing ground reaction forces emerge (*Santello, 2005*). However, conditioned reflexes are based and thereof depending on previous exposure and motor learning (*Santello, 2005*). In concern of an effective reflex utilisation, previous studies revealed that reflex delays and reflex gains inversely rely on the level of pre-activity, wherein elevated pre-activity increases reflex latency and gain (*Luoto et al., 1996; Hodges & Richardson, 1996; Stokes et al., 2000; Larivière et al., 2010*). Thus, it has been highlighted that muscular activity prior and post landing impact need to be *"regarded more a convention rather than a true physiological distinction"* (*Santello, 2005, p. 89*). Both collaborating predictive and reactive control mechanisms are complex and assumed to largely vary, depending on the given landing task constraints and previously acquired skill in the process of motor learning (*Santello, 2005*).

Uncertain Functional Trunk Conduct and Spinal Hazards.

Whilst predominate impact dissipation and absorption is undoubtedly implemented by the lower limbs, which previous landing research has mostly beheld; the true complexity of landing is believed to expand over a functional holistic body control (*Santello, 2005*). Such has unfortunately to date not been thoroughly investigated by previous landing studies, whereof the understanding of a trunk's integral involvement to the landing impact distribution can't be fully reckoned from available landing research. Few former studies have presented interrelations of the trunk's overall spatial orientation with lower limb joint angles (*Blackburn & Padua, 2008*) and muscular work (*Blackburn et al., 2009; Kulas et al., 2010*). Herein, each of these studies coherently suggested that stronger forward lean of the trunk would reduce stresses at the *ACL*, by forcing the overall landing performance into a less erect execution. Moreover, few authors suggested that an impaired neuromuscular trunk control and the presence of low back pain have a predictive influence on dynamic stabilisation of the trunk and the knee joint and may increase lower extremity injury risk (*Zazulak et al., 2007; Haddas et al., 2015, 2016a, b*). Contrastingly, *Prieske et al.* (*2013, 2015b*) concluded that trunk muscular activity does not substantially contribute to lower extremity stability and overall performance in drop landings and drop jumps. However, *Popovic & Kulig* (*2012*) conversely showed a compensatory elevation of trunk muscle activity, arising when lower extremity muscle deficiencies are present.

Apart of these revealed uncertain interrelations between the lower limb and trunk behaviour, only two previous studies have dedicated their investigations to the trunk muscles' functional remit for inherent spinal control and protection in landing executions (*Kulas et al., 2006; Iida et al., 2011*). *Iida et al.* (*2011*) herein suggested that the overall activity pattern of trunk flexor and extensor muscles favours the requisite trunk flexion attenuation rather than a co-contractive stiffening of the spine. Hereby *Iida et al.*'s conjecture (*2011*) stands in considerable contrast to conclusions of a large body of spine perturbation experiments, emphasising that trunk stiffening in preparation for and throughout an impinging impact is vital for spine protection (*Granata & Marras, 1995; Granata et al., 2001; Cresswell et al., 1994a; Krajcarski et al., 1999; Radebold et al., 2000; Santello, 2005; Hasan, 2005; Vera-Garcia et al., 2007; McCook et al., 2009; Shahvarpour et al., 2015*). This conception has, however, only been expounded for landing impacts in one sole landing study, which argued for a predominant implementation of abdominal bracing to stiffen the spine throughout landing (*Kulas et al., 2006*). Conversely, spinal stiffening in landings has been critically

appraised with regards to the hereof reinforced spine's exposition to augmented compression forces (*Recknagel & Witte, 1996*). The conflation of these previous arguments connotes a considerable complexity of demands posed to the spine, during landing executions. Herein it can be assumed that particularly the spine controlling trunk musculature is repeatedly challenged to provide accurate inherent spinal robustness against the landing evoked bending impulse (*Recknagel & Witte, 1996; Kulas et al., 2006; Iida et al., 2011*), while local eccentric muscular work must be concurrently provided to integrate the spinal joints in the absorption and dissipation of the landing impact (*Santello, 2005*). This supposition of the spine's incorporation in impact dissipation, has to date not been described by previous literature. Though, if the spine does not act as a single rigid element in the spring-like damping model (*Devita & Skelly, 1992; Recknagel & Witte, 1996*), and instead performs conjointly with the lower extremity on impact dissipation, spine segmental kinematic responses can be surmised. Herein, as landings universally involve a vertical impact load component, and have previously been shown to be most commonly performed with at least a minimal trunk forward lean, to evade spinal vertical compression forces from stiff erect landing (*Devita & Skelly, 1992; Recknagel & Witte, 1996*), largest bending load to the spine is hypothesised in the sagittal plane. Moreover, *Recknagel & Witte* (*1996*) emphasised the lumbo-pelvic area to be most vigorously affected by the landing impact, due to its proximate succession over the lower limbs and its inherent structural compliance; thus kinematic responses at this location are hypothesised to be substantially larger compared to more cranial spinal sections. Though, these hypotheses were hitherto not scientifically evidenced.
An above propounded conception of the trunk's remits in the control of landing impacts at the spine can conceivably be documented by an evaluation of kinematic spine segmental behaviour. Such observations could be moreover indicative for potential loading hazards to the spine. Considering the above described complex demands posed to the trunk musculature, an adversely large scope of failure is herein surmised. Hereby, abrupt spine segmental motion may depict inaccuracies in spinal robustness provided by the trunk musculature (*Reeves et al., 2007*), thus signifying hazardous spinal bending load onsets. Conversely, the absence of any spine segmental motion may thus indicate the presence of excessive spinal stiffening, leading to adversely large spinal compression forces (*Recknagel & Witte. 1996*), which are acknowledged as comparatively hazardous to the spine. Excessive sudden bending of the spine has been previously shown to potentially overload or injure the spinal tissue due to entailed precarious spinal shear (*Preuss & Fung, 2005*), which promotes the risk of low back disorder in occupational environments (*McGill & Norman, 1985; Marras & Mirka, 1990,*

1993; Marras et al., 1993; Norman et al., 1998). These findings on occupational loading factors might evenly apply for landing executions, as these studies consistently highlight that being subjected to high trunk acceleration demands contains a substantial risk for spinal overload; just as high linear trunk accelerations have been shown to constitute an integral part of landing (*Recknagel & Witte. 1996; Zhang et al., 2008; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*). Occupational studies moreover concerned a frequent exposition to sub-maximal trunk linear accelerations, as the hereof elicited trunk muscular co-activations providing spinal stiffness may fatigue, which would increase the risk of overexertion injuries to the spinal tissue (*Marras & Mirka, 1993; Magnusson et al., 1996; Cholewicki & McGill, 1996*). Though, potential spinal hazards in fatigued landing performers should be equally regarded in context of the presumed large frequencies, in which landings are often executed within single exercise sessions.

III - Hypothesised Spine Load Modifiers

Trunk Muscular activity.

Previous landing and experimental trunk perturbation studies have indicated that the most dominant trunk stabilising muscular activation pattern against an expected trunk antero-inferior impulse, such as induced by the vertical landing impact acting on the pelvis, forcing the trunk to forward flexion (*Iida et al., 2011*), involve sagittal lumbo-pelvic and dorso-ventral trunk muscular co-activations (*Lavender & Marras, 1995; Krajcarski et al., 1999; Radebold et al., 2000; Granata et al., 2001; Stokes et al., 2006; Lee et al., 2006; Kulas et al., 2006; Vera-Garcia et al., 2006, 2007; Mawston et al., 2007; Larivière et al., 2010; Popovich & Kulig, 2012; Haddas et al., 2016, a, b*). Within the majority of these experimental perturbation studies, trunk and pelvic stability was exposed to be most consistently provided by collaboration of musculus Erector Spinae, Obliquus Externus and Internus, and Rectus Abdominis (*Lavender & Marras, 1995; Krajcarski et al., 1999; Stokes et al., 2000, 2006; Mawston et al., 2007; Larivière et al., 2010*), whilst the most profound role in antero-inferior impulse counteraction was assigned to Erector Spinae (*Lavender & Marras, 1995; Krajcarski et al., 1999; Granata & Orishimo, 2001; Mawston et al., 2007*). These experimental studies, however, discordantly suggest that spinal robustness against trunk impulse perturbations either relies on preparatory muscular activity or muscular reactivity (*Granata et al., 2001; Lavender & Marras, 1995; Lariviere et al., 2010; Shahvarpour et al., 2015*). Though, it was shown that awareness of the perturbation characteristics lead to retained preparatory activity of trunk muscles and superior reactive muscular responses, reducing trunk excursions and velocities (*Lavender & Marras, 1995; Mawston et al., 2007*). Landing studies, however, have yet not conclusively revealed trunk muscular activity pattern stabilising the spine (*Kulas et al., 2006; Iida et al., 2011; Popovich & Kulig, 2012; Prieske et al., 2013, 2015b*), whilst a predominant feed-forward control of the trunk in landings is most conceivable in accordance to the evinced activity pattern of the lower extremity (*Santello, 2005*). Though, trunk impact characteristics in landing executions may emerge considerably unpredictable, due to a recognised large variability of lower extremity performances (*James et al., 2000; Nordin & Dufek, 2016; Nordin & Dufek, 2017; Nordin et al., 2017*), which might largely diminish the reliability of the pelvis reaching impact. This may, with regards to the previously acknowledged importance of impulse anticipation (*Lavender & Marras, 1995; Mawston et al., 2007*), critically hamper a meticulously concerted deployment of preparatory and reactive trunk muscular activity. Due to the hereof impeded provision of essentially preconfigured trunk muscular activity, more precarious sudden spine segmental bending can be surmised (*Lavender & Marras, 1995; Mawston et al., 2007*).

LANDING SKILL AND TASK DEMANDS.

Previous research has furthermore demonstrated disparities of trunk stabilising muscular activation pattern between genders (*Kulas et al., 2006; Lariviere et al., 2010*); whilst gender has concomitantly been shown to generally affect landing performances (*Decker et al., 2003; Pappas et al., 2007; Kernozek et al., 2008; Pappas & Carpes, 2012; Prieske et al., 2015a; Haddas et al., 2015; Weinhandl et al., 2010; Butler et al., 2013; Weltin et al., 2016*). These performance differences have been followed up to underlying disparities in landing skill (*Liederbach et al., 2008; Pappas et al., 2007, 2012; Pappas & Carpes, 2012; Bruton et al., 2013*). The effect of gender specific landing skills on landing performances has been moreover assumed to depend on the given landing task demands (*Weinhandl et al., 2010; Butler et al., 2013*). Harmonisation of landing skill and landing demands are generally believed to be most vital for efficient landing performances (*McKinley & Pedotti, 1992; McNitt-Gray, 2000; James et al., 2000; Santello, 2005*). Accordingly, motor task experience and skill have been frequently shown to capacitate individuals for superior muscular activity deployment and stabilisation of the spine (*Krajkarski et al., 1999; Skotte et al., 2004; Pederson et al., 2004, 2007; Mawston et al., 2007; Santos et al., 2010*). In the same light, landings under unfamiliar conditions (e.g. landing height elevations, extrinsically added trunk load) engender inferior lower limb performance (*McNitt-Gray, 1993; Zhang et al., 2000;*

Kulas et al., 2008; Nordin et al., 2016), which are presumptively ascribed to lacking motor adaptions to the task exigencies (James et al., 2000). Thus landing skill needs to be thoroughly considered as a significant modifier of landing performances regarding any neuro-motor activity involved. Unfortunately, a specific valuation of definable landing skills is difficult, and has thus yet not been expediently appraised in previous landing studies. Instead, several endeavours have been undertaken to reveal landing performance effects of different landing demands, which have herein mostly been investigated on generally landing familiar individuals (McNitt-Gray, 1993; Zhang et al., 2000; Santello et al., 2001; Liebermann & Goodman, 2007; Liederbach et al., 2008; Kulas et al., 2008, 2010; Weinhandl et al., 2010; Ambegaonkar et al., 2011; Janssen et al., 2012; Pappas et al., 2012; Dempsey et al., 2012; Dowling & Andriacchi, 2012; Prieske et al., 2013, 2015a, b). From these single-sided observations, it was shown that landing performances and impact magnitudes are substantially affected by the landing's overall gravitational potential energy, modified by landing height and performer's weight (Dufek & Bates, 1990; McNitt-Gray, 1993; Caster & Bates, 1995; Zhang et al., 2000; Kulas et al., 2008, 2010; Janssen et al., 2012). Moreover, investigations on experimentally distinctive landing types revealed that aggravations of limb support and landing surface compliance lead to overall stiffer landing performances under increased muscular activity, thus increasing kinetic loads to the lower extremity joints (Pappas et al., 2007, 2012; Weinhandl et al., 2010; Prieske et al., 2015a). Such effects, however, were occasionally shown to be diminished in landing performers being more familiar with the given landing condition (Liederbach et al., 2008; Weinhandl et al., 2010). Further task demand amendments, referring to practical landing executions in field sport disciplines, presented that an out of plane body extensions serving a ball catch in airtime lead to holistic landing performance alterations, which detrimentally raise lower extremity joint loads (Dempsey et al., 2012). By contrast to the most commonly revealed detrimental effects of landing demand augmentations, goal oriented elevation of musculoskeletal stiffness in drop jumps has been argued with its potential of alleviating lower limb bending loads, due to a herein enhanced conversion of the impact energy into the successive instant jump-off (Ambegaonkar et al., 2011; Hackney et al., 2016; Prieske et al., 2013, 2015b). Further experiments on landings under environmental concealment conditions moreover revealed significant dependencies on either visual or cognitive information for confident landing performances (Santello et al., 2001; Liebermann & Goodman, 2007).

Thus, alterations in landing task demands have conclusively been shown with common and meaningful effects to the lower extremity performance, which were former interpreted with altered joint loads and injury susceptibility. Yet, due to missing research, it is undetermined if such altered landing task demands would concomitantly alter the trunk's conduct and the hereof affected loading patterns to the spine. Such are, as based on previous research, assumed to significantly dependent on meticulously deployed trunk muscular co-activations, which adoptively rely on the performer's landing experience and motor skill.

IV - Methodological Consideration for Spine Load Quantification

Several load dimensions of landings have been previously assessed in landing literature. Herein few studies have to date commonly evaluated landing impact load to the trunk by means of linear surface accelerometry (*Recknagel & Witte, 1996; Ng et al., 2006; Zhang et al., 2008; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*). Even though linear accelerations can be measured in three singular dimensions, they do not facilitate an appraisal of angular bending load, which supposedly occurs alongside linear compression in each landing affected skeletal joint, including the spinal facets (*Recknagel & Witte, 1996; Kohrt et al., 2004; Santello, 2005*). Though, joint bending loads during landing have formerly been frequently assessed for the lower extremity. Herein measurements of ground reaction forces and inverse-dynamical calculations of joint impulses and peak joint moments for the ankle, knee, and hip were conducted on the basis of two- or three-dimensional angular joint motion data (*DeVita & Skelly, 1992; Zhang et al., 2000; Kulas et al., 2008, 2010*). Prejudicially, calculations of biomechanical inverse dynamic models have been critically appraised to rather depict close estimations of actual joint forces, since few factors (e.g. exact centre of mass position, mass distribution, rigidity and inertial moments of segments) in these calculations must be assumed, as they cannot be precisely measured (*Faber et al., 2016*).

Thus, few studies have evaluated joint loads in landings by means of mere angular velocities and acceleration derived from recorded joint motion data (*Siegmund et al., 2008; Popovich & Kulig, 2012; Iida et al., 2011, 2012*). Discrete kinematic measures have been shown to provide overall good to high within- and between-day reliability in the assessment of lower limb joint angular motion peak and waveform data in landings (*Ford et al., 2007; Millner et al., 2011; Malfait et al., 2014; Alenezi et al., 2014*), which was in most cases slightly superior to the reliability of the respective inverse-dynamically calculated joint kinetics (*Ford et al., 2007; Millner et al., 2011; Malfait et al., 2014*).

Though, few deviations in the reliability of plane specific motions were presented in those studies, which were discussed with the expended motions in these planes (*Ford et al., 2007*).

Previous studies have hereto argued that lower expanded motion ranges lead to relatively higher within- and between-day variability (*Kadaba et al., 1989; Ferber et al., 2002; Queen et al., 2006*). Comparatively low physiological motion ranges of the spinal facet joints (*Adams et al., 2006*) may thus considerably impair the overall reliability of kinematic spine segmental measures. In this regard, few previous research groups have investigated spine kinematics by means of simplified spine models, for which several sequent facet joints were clustered into adjacent v-shaped segments modelled over the course of the spine. Thusly constructed spine models were hitherto applied for spinal motion captures during dynamic sitting (*Preuss & Popovich, 2010*), lumbo-pelvic motion in unilateral landing (*Popovich & Kulig, 2012*), lifting tasks (*Mueller et al., 2016a*), gait (*Fernandes et al., 2016*), and stumbling (*Mueller et al., 2016b*) and have been shown to provide reliable outcomes (*Fernandes et al., 2016*), by confining the variability of herein measured spinal motions.
Yet, within-day variability of the hereof assessed motion captures has been shown to remain high in trunk responses to impact perturbations (*Mueller et al., 2017*). Landing performances have been moreover shown to be intrinsically largely variable within- and between-individuals (*Dufek & Bates. 1990; James et al., 2000; Fauth et al., 2010; Malfait et al., 2014; Nordin et al., 2016*). This was frequently shown for the lower limb segmental contributions to impact dissipation (*James et al., 2000; Santello. 2005; Nordin et al., 2016*). *James et al.* (*2000*) herein suggested that intra-individual variability alters non-linearly with added task demands (e.g. increasing landing heights). This presentation may portray neuromuscular system adaptions, leading to stable performances in landings, to which individuals were previously exposed, whilst unfamiliar landings may result in less consistent performances (*James et al., 2000*). Unfamiliarity of landings could potentiate the variability of spinal motion, since spine segmental motion is assumed to be largely affected by the impact remains of the lower limb shock absorption. Hence unfamiliar landings may induce largely unforeseeable perturbation impacts to the spine, altering within each trials lower limb performance.

In view of the acknowledged methodological and physiological boundaries, the assessment of spine segmental angular accelerations, derived from three-dimensional kinematics, may facilitate an in-vivo evaluation of striking spinal bending load during landing. Though concededly, without consulting inverse dynamic calculations and a neglect of segmental mass distributions, actual spinal load can hereby not be portrayed. However, measures of peak accelerations are hypothesised to illustrate the incidence of the most energetic stretch to the spinal tissue. Moreover, comparisons of angular acceleration within and between anthropometrically comparable individuals can be utilised as a discriminative parameter in landing and subject comparisons. By those means, peak spine segmental angular accelerations facilitate a verification of several yielded hypotheses on the onsets of sudden spinal bending load in landings.

RESEARCH OBJECTIVES

Based on the introduced previous literature findings and the herein clinically relevant desire of approaching an understanding for spine segmental bending load aspects in landings, this study pursues the following research objectives:

- To evaluate sudden spinal bending loads, by means of validly measured peak spine segmental angular accelerations, during various distinct drop landing tasks.
- To expose the spine location and motion plane experiencing highest peak spine segmental angular accelerations.
- To disclose and evaluate trunk muscular activity during drop landing and its influence on peak lumbo-pelvic flexion accelerations.
- To show alterations of drop landing induced peak lumbo-pelvic flexion accelerations and muscular activity in dependence on landing skill and landing task demands.

SCIENTIFIC HYPOTHESES

In pursuit of this study's research aims the following scientific hypothesis were postulated:

H1. Onsets of sudden spine bending loads during drop landings can be objectified and reliably assessed by peak spine segmental angular accelerations derived from a 3D motion captured spine model.

H2. Highest peak spine segmental angular acceleration in bilateral drop landing occurs in sagittal bending at the lumbar region.

H3. Drop landing induced peak lumbo-pelvic flexion accelerations are counteracted by conjunctional feed-forward and feed-back activation of spine stabilising trunk muscles.

H3.1 Peak lumbo-pelvic flexion accelerations are lower in previously landing familiar and male individuals as in landings followed by a premediated instantaneous follow-up movement, due to referred group specific concomitant adaptions in muscular trunk stabilisation pattern.

METHODS

RESEARCH DESIGN

In order to consecutively test this project's scientific hypothesis, two research trials were conducted in succession (*Figure 1*). The first study herein was carried out in a test-retest design (*M1, M2*) to investigate if peak spine segmental angular accelerations, measured by 3D motion capture, present reasonable and reliable outcomes within- and between-subjects and days (***H1.***). Furthermore data records from the first measurement session (*M1*) were evaluated to test if highest peak spine segmental angular accelerations will be encountered in sagittal flexion of the lumbar spine (***H2.***). Based upon the findings of our first research trial; a second research trial was implemented to investigate how trunk muscular activity counteracts peak spine segmental angular accelerations, by a multi-muscular composition and by each muscles phase specific attribution to neuro-muscular feed-forward and feed-back control (***H3.***). In a multi-group cross-sectional drop landing design, both, peak accelerations and muscular activity, were moreover tested for hypothesised differences between groups of presumptively disparate landing skill, emerging from contrasted landing familiarity and gender, on landing types with altered task demands (***H3.1.***).

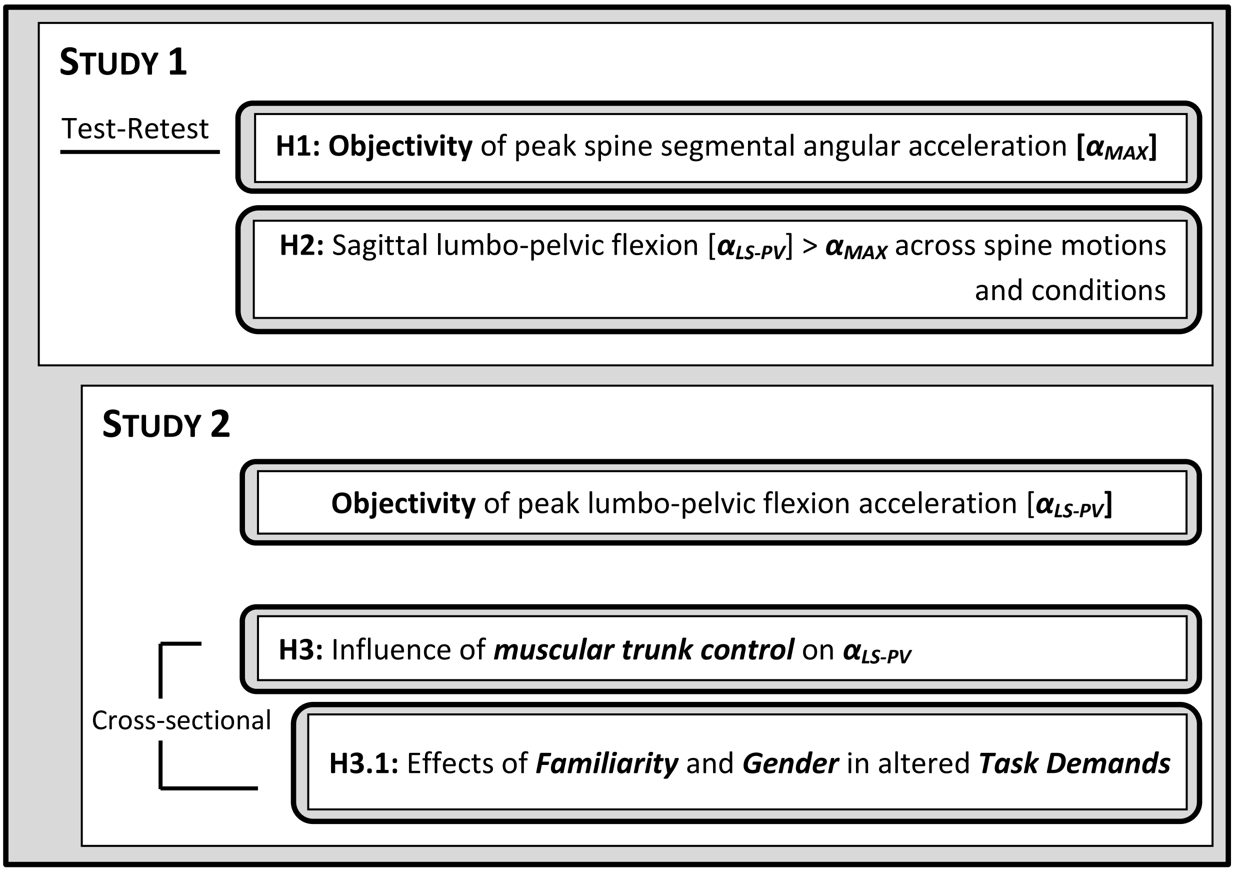


Figure 1: Research Design: Studies and Hypothesis (**H**)

Implementation of the research project was approved by the local Human Research Ethics Committee of the University of Potsdam.

STUDY 1

In order to evaluate the reliability and peak occurrences of sudden spine bending loads in drop landings, a multi-day test-retest study, with a wash-out period of at least 7 days between tests, was carried out in the University Outpatient Clinic - Center of Sports Medicine, Department Sports & Health Sciences at the University of Potsdam, Germany.

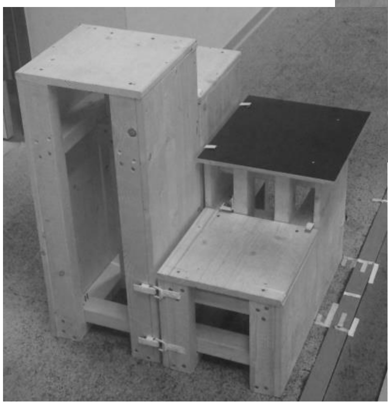
S1 - SUBJECTS.

17 healthy subjects (*n♂ = 9, n♀ = 8; age = 28.9 ±3.6yrs, height = 1.73 ±0.11m, weight = 69.6 ±13.4 kg*) with diverse involvement in physical activity (*Table A1.1*) were randomly recruited for the study. Health status of subjects was defined as: being free of any current neurological and physical complaints or impairments, the absence of any recent trauma or injury to the back and lower limb joints. These criteria were confirmed by a medical doctor in the University Outpatient Clinic to ensure the subjects' suitability and their capability of enduring the physical demands of the landing protocol. All subjects signed a written informed consent.

S1 - LANDING PROTOCOL.

In preparation for the execution of the drop landing protocol: subjects were asked to warm up by doing a short stair-run of 10 quick ascends and descends over 15 steps each, followed by 4 consecutive countermovement jumps. The subsequent protocol was comprised by a series of drop landing *Types* and *Heights* executed in substantially randomised order. Due to subject safety concerns, the protocol universally started with least demanding controlled bilateral drop landing [*BL*]. The following three *Types* were then fully randomised; comprised by unilateral landings [*UL*], fully blindfolded landing [*BF*], and landings involving an overhead catch of a ball during the dropping phase [*OH*]. The herein used ball (*10cm* of diameter) was hanging from the ceiling right above the marked landing area (*Figure 1.1*). Each landing type was previously described to the subject, with minimal instruction to the subject's individual execution providing that: the drop-off was performed bipedally and held to a minimal rise; the landing happened within the marked landing area and could be balanced instantly; the ball in *OH* was caught firmly. Each *Type* was performed "en bloc" from various *Heights* in random order. Landing *Heights* were practically predefined by a custom made stair-box, providing platforms to drop-off from *15cm*, *30cm*, *45cm*, and *60cm* above ground (*Figure 1.1*). The hereby applied variations of drop landings allowed the evaluation of trunk responses to various impact magnitudes, resulting from landing height and limb support alterations (*BL* vs. *UL*).

Unilateral landings were administered to generate an asymmetric impact introduction to the coronal trunk plane; whilst blindfolded landing led to full occlusion of visual-cognition of the landing surface, which was believed to impede physiological control over the required impact damping. Similar to blindfolding: the task, of catching a ball overhead within flight time, was believed to reduce the subjects' focus onto the landing execution, and hence restraining their landing performance. This task additionally impelled subjects to erect their trunk posture, by raising their head and arms towards the above mounted ball.



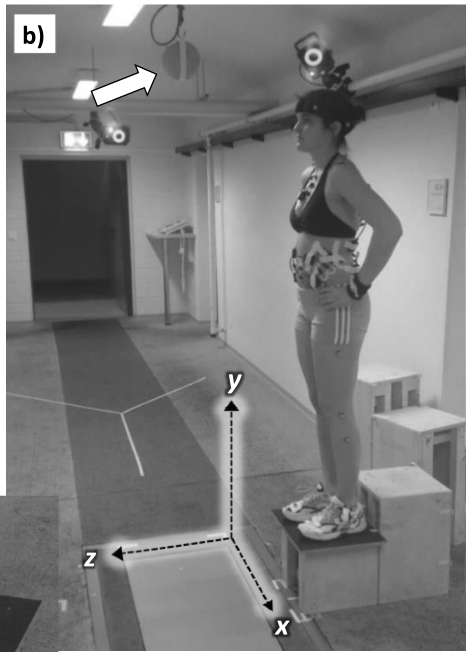


Figure 1.1: Experimental setup comprising 3D-Motion Capture System ('10 Cameras'; global coordinate system), Force Plate, Stair Box (*15, 30, 45, 60cm*), **b)** Ball hanging for *OH*

With regard to an assumed elevated risk of injury from landing impacts of larger heights, height restrictions were established; for females, which thus did not perform any drop landing from the *60cm* platform; and for *BF*, wherein merely landings from *15cm* and *30cm* heights were administered to all subjects. Moreover, subjects were allowed to withdraw from any landing variation, which they didn't have confidence in performing it safely. The basic landing protocol resulted in a minimum of 64 drop landings for males and 48 drop landings for females. Any attempt, failing due to a subject's violation of the landing execution instructions or due to equipment errors was repeated, until four valid trials from each landing *Type*Height* combination could be taken into analysis.

Between each block of *Type*Height*, a short resting periods for recreation was prescribed by one minute, whereas a resting period after each landing *Type* was set to three minutes in order to ensure fair tissue recovery between each block.

S1 - Equipment for Kinematic & Kinetic Measurements.

In order to analyse the trunk segmental behaviour, three-dimensional motion analysis, configured to a recording rate of *1000Hz*, was utilised (*3D video motion analysis system;* 9 cameras - *T10S; Vicon Motion Systems Ltd., Oxford, United Kingdom*). The system was deployed to capture the motions of a customised segmental spine model, previously used by *Mueller et al.* (*2016a, b*), leading back to *Preuss & Popovich* (*2010*), (*Figure 1.2*). This model allowed for depiction of the spine and the pelvis in four v-shaped adjacent 3dimensional segments, which were optically tracked over the whole course of landing: from the moment were the subject was told to initiate the drop off until the landing was fully stabilised.

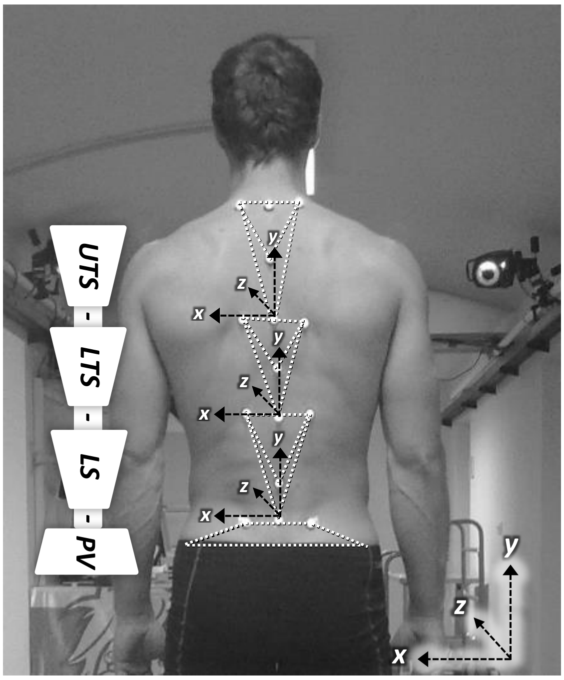


Figure 1.2: Spine Segment Model (adapted from *Preuss & Popovich* (*2010*)), comprising upper thoracic, lower thoracic, lumbar and pelvis segment (*UTS, LTS, LS, PV*); adjacent segmental joint motion captured in local 3dimensional coordinate systems originating from the upper spinous processus of the respective caudal segment (*UTS-LTS, LTS-LS, LS-PV*), the ground floor for *PV-GF*.

The segmental model was created by frames of reflective markers, which were preliminary to the landing protocol placed on superficially palpatable or estimated bony vertebrae and pelvis structures. Markers were attached using two sided duct tape, ensuring their hold to the skin. Hereto, the pelvis [*PV*], as the lowest segment of the construct, was modelled by a total of four markers placed onto the left and right bilateral anterior and posterior spina iliaca [*ASI, PSI*]. The cranially following adjacent spine segments modelling the lumbar- [*LS*], the lower thoracic- [*LTS*] and the upper thoracic spine [*UTS*] were each formed by 5 markers; whereas the lowest marker herein was respectively constituting the joint centre between two adjacent segments. *LS* was modelled by the spinous processi of *S1*, *L3* and *Th12* and the pairwise attendant *Th12* transverse processi. *LTS* was framed by one marker on the spinous processi of *Th12*, one *Th9,* one on *Th6* and 2 pair-wise markers on the *Th6* transverse processi. *UTS* was formed by markers on the spinous processi of *Th6*, *Th3* and *C7* together with the pairwise transverse processi of *C7* (*Figure 1.2*). Since transverse processi could not be manually palpated by the investigator, markers were applied to a close estimate over the targeted structure, by the use of a geometry triangle, applying these pairwise markers on a horizontal line with exactly five centimetre distance from the spinous processi of the respective vertebrae.

Kinetic ground reaction force [*GRF*] data and its timing was measured by a force plate, visibly integrated in the floor and concurrently representing the landing area (*AMTI Inc., Watertown, Massachusetts, USA*). Kinematic and kinetic data was simultaneously recorded by *"Vicon NEXUS 1.8©, Vicon Motion Systems Ltd., Oxford, United Kingdom"* and corporately fed to a personal computer. Anthropometric data of the subjects was measured by height and weight scales (*SECA 213; SECA 862*, *Hamburg, Germany*) on the first day of measurement (*Table A1.1*).

S1 - Data Analysis.

Processing. In order to observe merely the drop landing period from the taken recordings, only kinematic and kinetic parameters presented between *100ms* before initial ground contact [*IGC*], until the centre of mass reached its minimal vertical position [COM_{MIN}] were analysed. Within this timeframe kinematic data was processed applying semi-automated subroutines of "*Vicon NEXUS 2.1*" (*Vicon Motion Systems Ltd., Oxford, United Kingdom*). Herein the recorded markers were automatically labelled to the required above defined segments. In case this routine generated discernible incomplete or false outputs, manual correction of marker assignment had to be implemented by the investigator. Merely kinematic captures of trials with apparent appropriately modelled segments over the whole course of capture were further processed. Approved motion data was subsequently smoothed by a Woltring filter subroutine, fitted to the employed kinematic model and exported to an *ASCII* file, containing kinematic raw data on an accuracy of the *6th* decimal place of millimetre at a sampling rate of *1000Hz*. Respective crude kinetic force plate measures were also recorded with *"Vicon NEXUS 2.1"* at an equal sampling rate of *1000Hz*. Ground reaction force measures remained untreated and subjoined into the trial specific *ASCII* file which was subsequently exported to a spreadsheet software (*Microsoft Excel 2010®, Microsoft Corporation, Redmond, United States of America*) and herein further parametrised.

PARAMETRISATION. On the basis of waveform segment motion data over the whole period of landing, peak segmental angular acceleration magnitudes [α_{MAX}] and their respective timing [t-α_{MAX}] were calculated on the expression of *rad/s*^{2} and *ms* (*Figure 1.3*).

$$\boldsymbol{\alpha} = \frac{\boldsymbol{\rho}}{\boldsymbol{t^2}}$$

- *α = angular acceleration*
- *ρ = angle of rotation*
- *t = time*

Computation of α_{MAX} *and* t-α_{MAX} was conducted for three dimensionally segregated planes of *Motion*: sagittal, coronal, and transversal at each intersegmental angle between the pelvis to the ground floor [*PV-GF*] and at each of the 3 cranially following modelled spinal *Joints* (lumbo-pelvic [*LS-PV*], thoraco-lumbar [*LTS-LS*], thoracal [*UTS-LTS*]. Based on the global and local coordinate systems, predetermined by the *"Vicon NEXUS 2.1"* output, peak angular decelerations of the modelled segmental joints were chosen to describe sudden *Joint* angle reductions in the respective motion plane. The hereof described trunk loading events were sudden sagittal [*FLEX*] and coronal bending [*LAT*], and transverse twisting [*ROT*], (*Figure 1.2*). Notably, peak decelerations in each motion plane have by previous pilot tests shown to emerge on similar absolute scales of acceleration compared to their countermotion (*Figure 1.3*). In order to facilitate the presentation of the outcomes, peak segmental joint decelerations were expressed as the absolute magnitude of α_{MAX} in the *SI* unit radians per square second ($1rad/s^2 = 57.296°/s^2$).
The record of t-α_{MAX} in this study was merely employed to display the reasonable succession of α_{MAX} at each modelled spine *Joint* and *Motion* consecutive to *IGC*. Peak vertical ground reaction force in this study was expressed as an absolute measure by the unit of Newton [GRF_N].

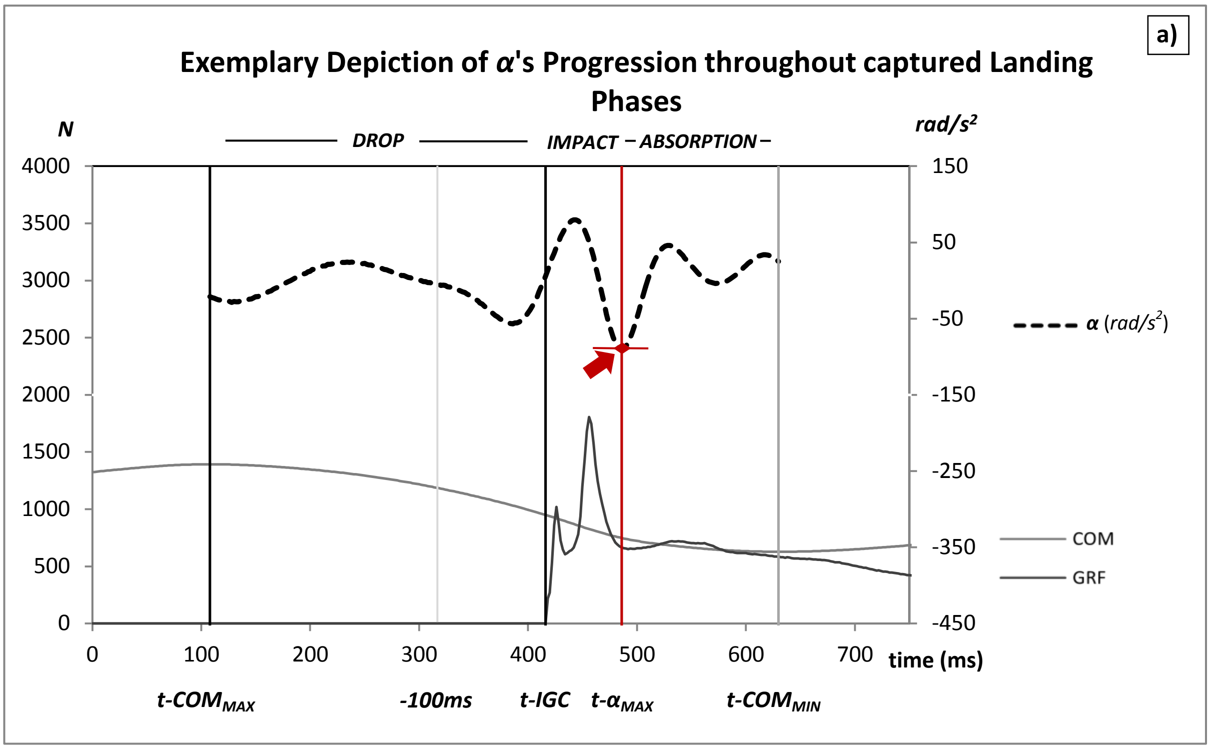
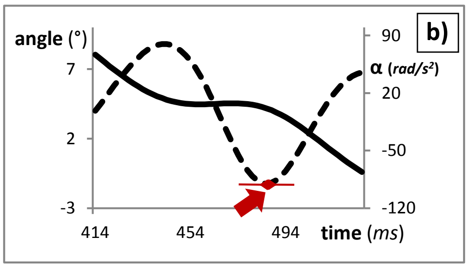


Figure 1.3 **a)** Parameter determination within course of landing:

- calculated temporal derivate of angular acceleration *[α](rad/s²)* from captured segmental joint angular motion (e.g. *FLEX/EXT$_{[LS\text{-}PV]}$*)
- determined landing phases (*DROP, IMPACT, ABSORPTION*) respectively delineated by peak centre of mass (*t-COM$_{MAX}$*), initial ground contact (*t-IGC*), peak segmental angular deceleration (*t-α$_{MAX}$*) and lowest *COM* position (*t-COM$_{MIN}$*) occurrences



b) scaled *LS-PV* angular change over time and resultant *$\alpha_{FLEX[LS\text{-}PV]}$ (rad/s²)*

Statistics. All statistical analyses were conducted using *SPSS* statistical software (*Version 23.0; SPSS, Inc., Chicago, IL, United States of America*). Each calculated inferential statistic was implemented on a set alpha level of ($\alpha=0.05$).

H1. To verify peak angular accelerations' sensitivity on the evaluation of landing's ground impact induced spinal bending load, α_{MAX} was initially checked for its ground impact affiliation. Hereto each trial's individual α_{MAX} was dichotomously categorised as reasonable when occurring consecutive to *t-IGC* [$t\text{-}IGC<t\text{-}\alpha_{MAX}$] or unreasonable if not. The proportion of reasonable α_{MAX} was expressed as a % ratio from all α_{MAX} records within particular landing *Types, Heights* and spine *Joints, Motions*, as for the overall aggregate of every α_{MAX} record. Furthermore $t\text{-}IGC<t\text{-}\alpha_{MAX}$ (%) was tested for rank associations to α_{MAX} using *Spearman's*

rank order correlation coefficient [r_S] once for aggregated and for each single constituent parameter.

Further objectivity assessments on α_{MAX} were performed by intra- and inter-individual within-day and between-days reliability analysis. Hereto, reliability analyses of α_{MAX} at diversified spine *Joints* and *Motions,* as at different landing *Types* and *Heights* were conducted. Intra- and inter-individual variability of α_{MAX} magnitude was statistically assessed by *Coefficient of Variation* for subject single measures (intra-individual [*CV%*]) and subject averages (inter-individual [$CV_{AV}\%$]) taken on the first day of measurement.

$$\boldsymbol{CV}\% = \frac{\boldsymbol{SD}}{|\boldsymbol{Mean}|}\%$$

Between-days reliability analysis was undertaken for α_{MAX}, by calculating descriptive *Means ±SDs* from *M1* and *M2*, as overall ranges of expressions from both measurement days [*Min., Max. (M1&M2)*]. Furthermore between-day measures were statistically assessed by *Intra-class Correlation Coefficient* [*ICC 2.1*], *Test-Retest Variability* [*TRV%*] and *Bland & Altman Analysis* [*Bias ± LoA*].

$$\boldsymbol{ICC(2.1)} = \frac{\boldsymbol{Var_b} - \boldsymbol{Var_{err}}}{\boldsymbol{Var_b} + (\boldsymbol{k}-\boldsymbol{1})\boldsymbol{Var_{err}} + \dfrac{\boldsymbol{k}(\boldsymbol{Var_{bk}} - \boldsymbol{Var_{err}})}{\boldsymbol{n}}}$$

- *k = number of rates*
- *n = number of cases*
- *Var_b = variance between cases*
- *Var_{err} = residual variance*
- *Var_{bk} = variance between rates*

$$\boldsymbol{TRV}\% = \left|\frac{\boldsymbol{M2} - \boldsymbol{M1}}{\boldsymbol{Mean(M1, M2)}}\right|\%$$

Bland & Altman Analysis *Cartesian coordinates:* $\boldsymbol{M(x,y)} = \left(\frac{\boldsymbol{M1+M2}}{\boldsymbol{2}}, \boldsymbol{M2} - \boldsymbol{M1}\right)$

$$\boldsymbol{Bias} = \frac{\boldsymbol{1}}{\boldsymbol{n}} \cdot \sum(\boldsymbol{M2} - \boldsymbol{M1})$$

$$\boldsymbol{LoA} = \boldsymbol{Bias} \pm \boldsymbol{1.96\ SD}$$

In order to reference α_{MAX}'s overall reliability and reasonability to those of the accompanied ground reaction forces, intra- and inter-individual within (*CV%; CV_{AV}%*) and between-days reliability (*ICC 2.1, TRV%, Bias ± LoA*) was additionally analysed for GRF_N.

H2. To test the hypothesis that highest accelerations would emerge in flexion of the lumbo-pelvic joint, $\alpha_{FLEX[LS\text{-}PV]}$ was compared to α_{MAX} of other spinal *Joints* and *Motions*. Hereto *Means* ± *SDs* of $\alpha_{FLEX[LS\text{-}PV]}$ and α_{MAX} from individual subjects and from the overall cohort were assessed and compared for each spinal *Joint**Motion* on single and the overall average of all landing *Types & Heights*. Each descriptive group analysis was followed by *Independent Samples T-test*. To furthermore reveal the influence of various vertical landing conditions on peak lumbo-pelvic flexion accelerations, $\alpha_{FLEX[LS\text{-}PV]}$ from each assessed particular landing *Type*Height* were compared with each other. To further reveal magnitude relations between peak ground force and spine peak segmental angular acceleration, GRF_N was tested on its linear associations to α_{MAX} overall and in each spinal *Joint* and *Motion*, using *Pearson's Correlation Coefficient* [*r*]. For each herein conducted analysis only data recorded in *M1* was used.

STUDY 2

To investigate the onset of peak lumbo-pelvic flexion accelerations in dependence on trunk muscular activity during drop landing, a multi-group cross-sectional study was carried out in the University Outpatient Clinic - Center of Sports Medicine, Department Sports & Health Sciences at the University of Potsdam, Germany.

S2 - SUBJECT ENROLMENT.

To implement the multi group design of this study, initially 45 healthy subjects (24♀***F***, 21♂***M***) were recruited into groups with distinct landing *Familiarity*. Health status of subjects was scrutinised and feasible if subjects were free of any current neurological and physical complaints or impairments, in the absence of any recent trauma or injury to the back and lower limb joints. These criteria were confirmed by an anamnesis questionnaire, which was filled by the lead investigator during a face-to-face interview with the subject or during a phone call prior to the measurement. Further general inclusion criteria, enquired within the prior anamnesis, presupposed subjects to be normal weight and of age between 18 and 40 years. Normal weight by means of *Body Mass Index* ([*BMI*]: *♀19-24, ♂20-25*) was scrutinised by the calculation of anthropometric measures, taken by height and weight scales (*SECA 213, SECA 862, Hamburg, Germany*)

The multi group design required to distinguish *Female* as *Male* subjects by their previous and or current jumping and herewith accompanied landing experience. The approach of discriminating subjects by landing *Familiarity* was established to identify individuals with practised landing skills, which are assumed to be compound by better lower extremity shock absorption and improved spine stabilisation performance (*Santello, 2005*). Hereto subjects were asked about their retrospective lifetime jumping/landing exposure whereof two premediated groups of landing *Familiarity* were composed: The first group included individuals which reported current or preceding participation in sport- or training types involving jumping activities within their past 10 years and an accumulated quantity of more than 1000 landings in that timeframe [*Familiar ≙ 1*]. To disclose the total amount of landings per individual, each reported activity, period of participation and a close estimate of jumps per week were queried. Herein respondents were briefed that solely jumps with limb supported landings on solid ground were relevant for the survey. This criterion was relevant to exclude jumps reported from disciplines like high jump, pole vault, high diving or horseback riding, etc.. To sharply distinguish two groups of landing *Familiarity*, the contrasted second group included only individuals which, upon the survey, accumulated less than 100 foot first landings in their past 10 years [*Unfamiliar ≙ 2*]. Individuals with landing exposure scoring between the 2 delimiting set points were rejected from study participation. This discriminating directive resulted in a total recruitment of 25 landing *Familiar* subjects (13♀*F1*, 12♂*M1*) and 20 landing *Unfamiliar* subjects (11♀*F2*, 9♂*M2*), registered in this study. All subjects signed a written informed consent.

S2 - Breaking Test and Landing Protocol.

Before undergoing the landing protocol, subjects had to complete five manually assisted muscular breaking tests, modelled after recommendations by *Konrad* (*2005*), targeting different trunk muscles. Breaking tests were conducted to identify the respective muscles individual maximal voluntary contraction capacity [*MVC*] for later normalisation of muscle individual electromyography [*EMG*] measures during landing. Each breaking test was executed in the middle position of the targeted muscle's range of motion [*ROM*] with the individual's maximal contraction effort held for five continuous seconds against the respective muscle loading force. *MVC* muscle tests were performed for M. Rectus Abdominis [*RA*], M. Transversus Abdominis & Internus Abdominis [*TrA-IO*], bilateral M. externus abdominis [*EO*] and M. Erector Spinae [*ES*]. *RA* was tested for sagittal flexion torque production in a seated trunk reclined position with a posteriorly tilted pelvis and feet positioned on even ground. Movement from the initially adjusted middle *ROM* position was inhibited by the investigator persisting the required testing time. Hereto the investigator placed his one forearm against the thighs and his second forearm against the shoulder girdle of the subject, pushing with counteracting proportionate force against the subject's produced muscle torque. *TrA-IO* was tested in an *RA*-test similar seated position, with the alteration of the subject rotating its shoulder girdle antero-laterally around the spines longitudinal axis over the pelvis away from the targeted *TrA-IO* muscle side. The manually added resistance by the investigator herein counteracted the tested subject's flexion and rotation torque. Bilateral breaking tests for *EO* were initiated from a prolate lateral decubitus position, whereat the subject was required to hold a slightly upwards tilted lateral plank exercise with straightened legs and arms crossed over the thorax; such as only the unilateral hip and gluteus maximus area were used as ground support. Torque by the subject was produced medio-laterally in the coronal plane along with an anterolateral trunk rotation around the spines longitudinal axis. Resistance by the investigator was provided with one hand pushing the outer thigh downwards and the second hand pushing the upper positioned shoulder of the subject down- and backwards (postero-lateral flexion rotation). Latter, bilateral *ES* was tested on a quasi-prone positioned subject with the pelvis and thighs lying on a padded crate and supportively held stable by the investigator's continuous forceful downward push. The tested subject was meanwhile impelled to hold its whole upper body without ground support horizontally, by producing peak sagittal extension torque.

Ahead of the drop landing protocol execution, subjects were asked to warm up by exerting a short stair-run of 10 quick ascends and descends over 15 steps each. Following the warm up, a brief instructed familiarisation with the subsequently desired landing standardisation was prescribed to each subject. Familiarisation was carried out by unrecorded two countermovement jumps and three consecutive drop jumps on the even ground of the landing area, onto which later landings were executed. Subjects were herein instructed to attach their hands to the lateral pelvic area surface (hands akimbo), right under the iliac crest, remaining on spot throughout the whole jump and the subsequent landings.

The landing protocol included multiple (*8 to 11*) repetitions of three partially randomised predominantly vertical bilateral drop landings from a standard height platform of 45cm. Standardisation of landing execution was set to the kept fixed hands akimbo position, a bilateral take-off from the drop platform and a respective bilateral landing onto the ground landing area. Furthermore, a minimal rise when jumping from the platform was compulsory in order to reduce any addition of dropping height. The first two landing *Types* differed by the instantaneous follow up tasks and were administered in random order. These landings comprised: pure drop landings [*DL*] without any follow up movement and drop jumps [*DJ*]. Drop jumps were described as performing a subsequent instant vertically directed jump-off, under minimal ground contact time spend. However, it was herein not necessary for subjects to perform a plyometric drop jump reaching *<200ms* ground contact time to succeed in a trial, as otherwise referenced in literature (*Ball et al., 2010*). Both *DL* and *DJ* consisted of 8 valid trial repetitions each.

The third landing *Type* differed from the first two by predictability of the follow up motion [*DS*]. Hereby subjects were only prompted to either perform a *DL* or a *DJ* subsequently to their take-off from the landing platform. The hereby initially concealed randomised follow up movement was triggered by a strong light signal of a standard lamp (*600 Lumen*) positioned in the subjects field of vision. The decisive light signal, prompting a *DJ* execution, was released after the subject's feet dropped below the platform height after take-off; the activation of the light signal was manually controlled with a remote switch, held by the investigator. *DS* landings were performed until 11 valid trials were recorded. For each landing *Type* only the drop landing from the platform onto the ground was of interest for the study aim and thereby taken to analysis. Any recorded attempt, which was distorted during the measurement by either equipment or execution failure, was repeated until an at most third additional total repetition per landing *Type* was conducted. The number of additional repetitions was hereby restricted to avoid confounding fatigue of subjects, which was otherwise controlled by resting periods of three minutes between each landing *Type*. For each landing *Type* at least one test trial before recording was granted to the subjects. With the intention of avoiding landing execution pattern copied from the investigator, which has been undesirably observed in pilot tests, depiction of landing *Types* was not undertaken for *DL* and *DS*. Only *DJ* was demonstrated to the subjects to ensure the comprehension of the immediacy of the follow up jump.

S2 - EQUIPMENT AND SUBJECT PREPARATION.

KINEMATIC & KINETIC DEVICES. Three dimensional motion analysis (*3D video motion analysis system, 8 cameras - T10S, Vicon NEXUS 2.1©, Vicon Motion Systems Ltd., Oxford, United Kingdom)*, set to a recording rate of *500Hz*, was utilised capturing the lumbo-pelvic region. The in this study used model (*Figure 2.1*) conformed to the lumbar and pelvic segment from the previously employed spine segmental model (*Figure 1.2*). Hereto 11 markers were placed by one experienced investigator on the subject's skin, using two sided duct tape. The pelvis [*PV*] was modelled by six markers (*10mm* of diameter). Four were placed onto the left and right bilateral posterior and anterior spina iliaca [*PSI, ASI*] and one additional marker on each side of the waist centred between the respective *PSI* and *ASI*. The waist markers herein were applied as bilateral *ASI* backup reference markers, necessary for later software implemented reconstruction of the pelvis, in case a large forward bend of the trunk during landing obscured the *ASI*. The cranially adjacent lumbar segment [*LS*] was formed by five markers (*5mm* of diameter) placed on spinous processi of *S1, L3, Th12* and *Th12*'s pairwise attendant transverse processi. Since transverse processi could not be manually palpated by the investigator, markers were applied to a close estimate over the targeted structure. Hereto an assisting geometry triangle was

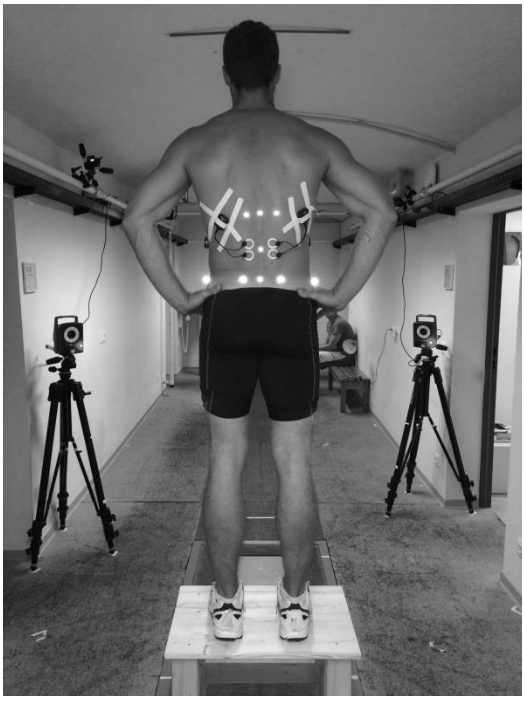


Figure 2.1: Lumbo-Pelvic Marker Model

used to draw a horizontal line over *TH12*'s location, onto which transverse processi were marked at exactly five centimetre distance from the spinous processi. The modelled segments were visually tracked over the whole course of landing: from the moment were the subject was told to initiate the drop-off until the landing was fully stabilised.

Kinetic ground reaction force [*GRF*] data was measured by a force plate (*AMTI Inc., Watertown, Massachusetts, USA*) recording at *1000Hz*. The force plate was visibly integrated in the floor and represented the landing area. Kinematic and kinetic data was simultaneously recorded by "*VICON Nexus 2.1*" and corporately fed to a personal computer.

ELECTROMYOGRAPHY [EMG]. Surface *EMG* [*sEMG*] of six selected trunk muscles was recorded from each subject during their landing executions. Pairwise *SEMG* electrodes (*Ambu, Medicotest, Denmark*, type: P-00-S, centre to centre distance 25mm) were, according to muscle fibre directions, placed over the midpoint of each muscle belly. The therefore utilised anatomical bony insertion landmarks were previously determined by palpation. Preparations were performed for: *RA*-left (*3cm* lateral to the umbilicus), *EO*-left & -right (each side *15cm* lateral to the umbilicus), *TrA-IO*-left (centred to a straight line between superior iliac spine and symphysis pubis), and *ESL3*-left & -right (*3cm* laterally to *L3*). In order to provide stable electrode contact and low skin impedance before electrode placement, the subject's skin over the particular area, was prepared by a standardised protocol as follows: **1.** hair was removed using a razor blade, **2.** skin was slightly abraded with abrasive paper, **3.** skin area was cleaned, using an alcohol containing solution, wiped over with a paper tissue. This procedure produced pursued impedance levels below *5kΩ*, being considered a very good conductance condition (*Konrad. 2005*); otherwise all steps of skin preparation were repeated until the impedance fell below the aimed critical value. Electrodes were wired to portable *Myon320 Transmitters* (*Prophysics AG, Zuerich, Swizerland*), which were strapped to the subjects body, using skin compatible duct tape. The pre-amplified *EMG* signals, collected at a sampling frequency of *4000Hz*, were then Bluetooth-transmitted to a *Myon320* Receiver (*Prophysics AG, Zuerich, Swizerland*), before data was *A/D* converted and, collectively with the trigger signal of the force plate, recorded by "*IMAGO Record Master*" (*Pfitec® Biomedical Systems, Endingen, Germany*) and saved to a personal computer.

S2 - Data Analysis.

Processing. Kinematic data in this investigation was processed by the same routines as performed in the previous study (see "*S1 - Data Analysis - Processing*"). *SEMG* data of *MVC* containing muscle specific breaking test measures and landing trial *sEMG* data of all 6 trunk muscle leads were initially processed with "*IMAGO ProcessMaster v.5.43"* (*Pfitec® Biomedical Systems, Endingen, Germany*). Each recorded *EMG* channel per landing trial and *MVC*-recording was herein **1.** baseline offset corrected, **2.** 4^{th} order Butterworth band-pass filtered at *30 to 500Hz,* **3.** full wave rectified. Filtering of raw *EMG* recordings was implemented to debug records from signal noise and cardiac activity artefacts. For this purpose the lower cut-off frequency of *30Hz* was chosen to remove occurring *QRS-wave* contamination of the *EMG* signal (*Redfern et al., 1993; Drake & Callaghan, 2006; Butler et al., 2009; DeLuca, 2010*), while upper cut-off frequency of *500Hz* was applied to reasonably smooth the recorded stream, by excluding frequency spectra which are rather attributed to signal noise components than to actual *EMG* activity (*DeLuca, 2010*). The low pass frequency of *500Hz* in our filtering was hereby in agreement to *ISEK* and *SENIAM* recommendations. However high pass filter frequency recommendations couldn't be matched, since the *ECG* contamination removal required to exclude the spectra below *30Hz* (*Redfern et al., 1993*).

Purified *EMG* signals were subsequently exported to *ASCII*-Files and further processed by therefore specifically developed *MS. Excel* Macros (*Microsoft Excel 2010®, Microsoft Corporation, Redmond, United States of America*). Herein *EMG* signals were smoothed by a moving root mean square transformation [*RMS*], encompassing epochs of *20ms*.

Data Reduction & Parametrisation. All landing trial specific kinetic, kinematic and electro-myographic recordings were initially cut down to the period of landing; This was defined from peak rise height after take-off, determined by the recorded highest vertical *PSI* position, to the end of drop landing absorption, determined by the lowest vertical *PSI* position. Highest and lowest *PSI* positions were confidentially considered as representative markers for the concomitantly reached maximal and minimal vertical centre of mass position [COM_{MIN}]. Hereto a previous assessment has shown significant strong linear positive association between *PSI* and *COM* in magnitude (r =.98, p=.000) and time (r =.95, p=.000); with $PSI = .854 \cdot COM + 64ms$.

Peak sagittal lumbo-pelvic flexion accelerations around the highest sacral vertebrae (*S1*), representing the joint-centre between *LS* and *PV* [$\alpha_{FLEX[LS\text{-}PV]}$], were derived from the kinematic motion capture. By targeting exclusively landing impact induced peak lumbo-pelvic flexion accelerations, highest $\alpha_{FLEX[LS\text{-}PV]}$ occurring subsequent to the kinetically recorded initial ground contact [*t-IGC*] of each trial were taken into analysis. In cases where peak expressions of $\alpha_{FLEX[LS\text{-}PV]}$ appeared before *t-IGC*, the successive peak of $\alpha_{FLEX[LS\text{-}PV]}$ occurring post *t-IGC* was analysed. Each trial's individual *t-IGC* was hereby, deduced from *GRF* data and defined by a vertical force component onset above *10kg*.

In order to facilitate temporally distinct analysis of muscle activity, kinetic and kinematic load parameters were used to determine three distinct landing phases. The first phase herein, determined as the dropping phase [*DROP*], was delimited from the time of *PSI*'s peak rise height to the time of *IGC*. The subsequent phase [*IMPACT*] reached from *t-IGC* to the occurrence of $\alpha_{FLEX[LS\text{-}PV]}$. These two phases were for concomitant analysis additionally merged to a collective pre-phase [*PRE*]. The remaining time course of landing, until the minimal *PSI* position was reached [*ABSORPTION*], was with respect to the focus of this study rather subordinately regarded. (*Figure 1.3*)

SEMG-RMS from each specific muscle and landing trial was analysed on averages of each determined landing phase, expressed as absolute and *MVC* normalised measures. Hereto *MVC SEMG-RMS* waveform data from muscle specific breaking test was averaged over continuous three seconds of peak activation, revealing six muscle specific *MVC* measures per subject. *MVC* normalised landing trial muscular activity was expressed as percent ratio of maximal voluntary contraction [*%MVC*].

To supplementary expose the ratio of antagonistic co-activation between bilateral lumbar Erector Spinae and Externus Obliquus, co-contraction indices between [$EO_{l\&r}$] and [$ES_{l\&r}$] were calculated [*CCI[EO|ES]*] (*Lewek et al., 2004*) during *PRE*, *DROP* and *IMPACT* phase of landing, using the following formula:

$$\boldsymbol{CCI[EO|ES]} \;=\; \frac{\mathbf{1}}{\boldsymbol{n}}\left[\sum\nolimits_{\boldsymbol{i}=\mathbf{1}}^{\boldsymbol{n}}\left(\frac{\boldsymbol{EMG_{Small(i)}}}{\boldsymbol{EMG_{Large(i)}}}\right)\left(\boldsymbol{EMG_{Small(i)}}+\boldsymbol{EMG_{Large(i)}}\right)\right]$$

Co-contraction Index [*CCI*] Formula (*Lewek et al., 2004*)

For the presentation of single muscle activities, *sEMG* activity of bilaterally recorded muscles [$EO_{l\&r}$; $ES_{l\&r}$] were reduced to the data of the left muscle lead; both muscles showed no statistically significant (*sig.*) difference between their left and right side activity.

STATISTICS. Initially, subject group characteristics (*age, height* and *weight*) within the determined groups of landing *Familiarity* and *Gender* were descriptively compared and inferentially tested for statistically sig. differences, using *independent samples T-test*.

RE-VERIFICATION OF $\alpha_{FLEX[LS\text{-}PV]}$'S WITHIN-DAY RELIABILITY. In order to re-verify the reliability of peak lumbo-pelvic flexion acceleration in a recruited larger cohort and within dissimilar group characteristics: $\alpha_{FLEX[LS\text{-}PV]}$ was accordingly to the previous study checked for its sensitivity by means of $\alpha_{FLEX[LS\text{-}PV]}$'s ground impact affiliation [*t-IGC*$<$*t-*$\alpha_{FLEX[LS\text{-}PV]}$*(%)*], and its inter- and intra-individual within-day variability overall and within dissimilar groups of landing *Familiarity* (*1,2*) and *Gender* (*M, F*) as for disparate landing *Types* (*DL, DJ, DS*). Equally, inter- and intra-individual variability of trunk muscular *PRE*-activity was examined overall and in factorial groups of subjects and landings. Intra-individual within-day variability analysis was conducted by means of *Coefficient of Variation* [*CV*] of the individual subject dispersion within the 8 to 11 trials performed on each landing. Inter-individual variability was calculated on the averages of the task specific repetitions of each subject [CV_{AV}].

H3. Several statistical assessments were conducted, in order to test the multifaceted hypothesis that peak lumbo-pelvic flexion accelerations would be depending on trunk muscular pre-activity, which would moreover mutually alter with previous landing *Familiarity*, *Gender* and the implementation as the predictability of an instant follow-up task. Hereto initially $\alpha_{FLEX[LS\text{-}PV]}$ magnitudes between factorial groups of *Gender* (*M, F*) and *Familiarity* (*1, 2*) were compared descriptively within and between different landing *Types* (*DL, DJ, DS*). Averaged data of the assessed cohort and groups was descriptively presented by means and standard deviations [*Means ± SD*] and by parameter expression ranges [*MIN, MAX*] of single subject means within the cohort and within groups. Contrasts between compared groups were expressed as absolute and respective relative difference [*Diff.*; *%Diff.*]. To inferentially evaluate within- and between-subject effects of *Gender*, *Familiarity* and landing *Types*, on $\alpha_{FLEX[LS\text{-}PV]}$, a *Three-way mixed ANOVA* was conducted. This was followed with *Post-hoc Bonferroni test* for pairwise comparisons between single factors and factor groups.
In order to disclose the influence of muscular *PRE*-activity to $\alpha_{FLEX[LS\text{-}PV]}$, multiple statistical examinations were performed. Hereto, analogously to the implemented statistical analysis of $\alpha_{FLEX[LS\text{-}PV]}$, each muscular *PRE*-activity (*%MVC*) of *[EO]*, *[ES]*, *[RA]*, *[TrA-IO]* and *CCI[EO|ES]* within groups of *Gender* (*M, F*) and *Familiarity* (*1, 2*) were compared

descriptively within and between different landing *Types* (*DL, DJ, DS*). These comparisons were equally inferentially tested by *Three-way mixed ANOVA* and *Post-hoc Bonferroni tests.*
In order to moreover facilitate an interpretation of differentiated pre-mediated and reflexive muscular contribution counteracting $\alpha_{FLEX[LS\text{-}PV]}$, deployed muscular pre-activity (*%MVC*) of *[EO]*, *[ES], [RA],* and *[TrA-IO]* was discretely analysed for *DROP* and *IMPACT* phase. Additionally, each muscle's unnormalised total *sEMG-RMS* power contribution to the landing was segregated and expressed to its relative phase specific share [$\%_{LANDING}$] to the *DROP* and *IMPACT* phase. Muscle contributions and activations following $t\text{-}\alpha_{FLEX[LS\text{-}PV]}$ ($\triangleq$ *ABSORPTION*) were disregarded in this analysis. Both parameters ($\%_{LANDING}$, *%MVC*) for each muscle were subsequently inferentially tested for differences between landing phases, using *paired samples T-test*. Moreover, descriptive phase specific muscular contribution and activity differences between pairwise groups of landing *Familiarity* and *Gender* in each landing *Type* were furthermore inferentially tested by *independent samples T-test*.
Additionally to the statistical comparisons of $\alpha_{FLEX[LS\text{-}PV]}$ and trunk muscular pre-activity between factorial groups, each muscular activity (*[EO]*, *[ES], [RA]*, *[TrA-IO]* and *CCI[EO|ES]*) was tested for statistical associations to $\alpha_{FLEX[LS\text{-}PV]}$ overall and within each factorial group. These associations were discretely analysed for *%MVC* and $\%_{LANDING}$ activity in *PRE-*, *DROP* and *IMPACT* phase, using *Spearman's rank order correlation coefficient* [r_S] analysis.
Any implemented inferential statistic was conducted on a *5%* alpha level of significance ($\alpha = 0.05$). Strengths of found associations were labelled after the declaration by the British Medical Journal (*BMJ, London, United Kingdom*).

RESULTS

S1 - OBJECTIVITY OF α_{MAX}

SENSITIVITY.

In *≈91%* of *1200* landings measured, α_{MAX} occurred consecutive to *t-IGC*. The rate percentage of these detections slightly differed between observations in different constituent factors (*Table 1.1*). Within factors constituting, trends were shown for landings engendering higher impacts presenting higher *t-IGC<t-*α_{MAX} rates. These were most noticeable across increasing heights. Furthermore with less certainty, *t-IGC<t-*α_{MAX} analysis indicates higher affiliation rates in more caudal spinal *Joints*.

Table 1.1: Ground impact affiliation of α_{MAX} (*t-IGC<t-*α_{MAX} *(%)*) in each level of landing *Type & Height,* spinal *Joint & Motion*

TYPE	*BL* **90**	*BF* **88**	*OH* **90**	*UL* **96**
HEIGHT	*15* **88**	*30* **91**	*45* **93**	*60* **95**
JOINT	*UTS-LTS* **83**	*LTS-LS* **95**	*LS-PV* **92**	*PV-GF* **96**
MOTION	*FLEX* **91**	*LAT* **93**	*ROT* **90**	

Overall rank associations between α_{MAX} and *t-IGC<t-*α_{MAX}*(%)* were statistically sig. strong positive: $r_S(168)= .68$, $p= .000$ (*Figure 1.4*). Depth analysis within constituent factors showed predominate homogeneity of such positive associations in each factor level, despite *LTS-LS*: $r_S(42)= .11$, $p= .470$. (*Table A1.2*). In contrast α_{MAX} of *LS-PV* was pre-eminently very strongly positive associated to *t-IGC<t-*α_{MAX}*(%)* ($r_S(42)= .85$, $p= .000$).

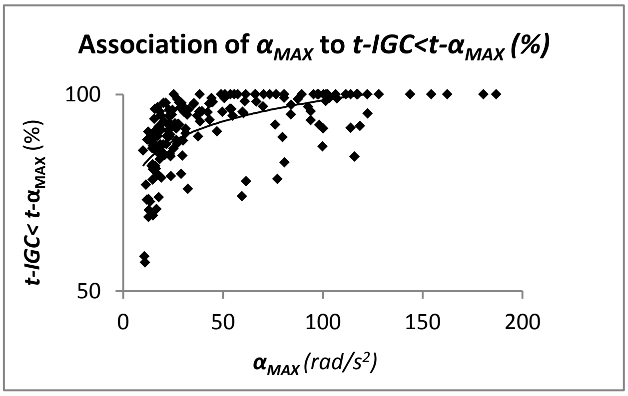


Figure 1.4: Association between α_{MAX} to *t-IGC<t-*α_{MAX} *(%)* plotted by α_{MAX} measures from each spine *Joint*Motion** landing *Type*Height* combination from subject averages.

WITHIN-DAY RELIABILITY (INTER- & INTRA-INDIVIDUAL).

Wide ranges of α_{MAX} measures were present across all landing *Types, Heights* and spine *Joints*, *Motions* (*Min: 1, Max: 936rad/s*^{2}). Overall inter-individual within-day variability of α_{MAX}, averaged across all landings and spine motions, was CV_{AV} *=66%*. Within landing *Types*Heights, BF30* showed highest variability (CV_{AV} *=89%*) while lowest variability between individuals was found on *UL60* (CV_{AV} *=42%*); notably *UL60* was solely performed by 6 subjects. On the observation of spinal *Joints & Motions, FLEX* presented an overall slightly lower variability than other *Motions* (*Table A1.3*). Intra-individual variability of α_{MAX} was, compared to inter-individual variability, on the average of all subjects (*CV =36%*) remarkably lower. None of the subjects showed significantly smaller or larger intra-individual variability in any particular landing or *Motion* (CV_{MIN} *=26%*, CV_{MAX} *=49%*), (*Table A1.4*). Compared to α_{MAX}, GRF_N across landing *Types*Heights* showed substantially lower inter-individual variability averaging at CV_{AV} *= 9%* ($CV_{AV\text{-}MIN}$ *=6%*, $CV_{AV\text{-}MAX}$ *=11%*). Meanwhile intra-individual variability of GRF_N was similarly low (CV_{MIN} *=4%*, CV_{MAX} *=12%*), thereby presenting around one third of variability found in α_{MAX}.

BETWEEN-DAY RELIABILITY.

Overall reliability of α_{MAX} was presented in margins from "poor" to "good" agreement between measurements (ICC_{MIN} *=.004,* ICC_{MAX} *=.811*; *Koo & Li, 2016*), with systematic errors of measurement (*Bias*) from *0* to *-29rad/s²* and variabilities between measurements (*TRV%*) of *24* to *56%*. Whilst hereby *TRV%* was only trivially divergent between distinct combinations of spinal *Joint*Motion* and landing *Type*Height*, *ICC* and *Bland & Altman Analysis* showed particular differences within and between spine *Joints*Motion* and landing *Type*Heights*.

SPINAL JOINTS*MOTIONS. Best agreement of α_{MAX} measurements between days was shown for *FLEX* in all *Joints* ($ICC_{UTS\text{-}LTS}$ *=.811,* $ICC_{LS\text{-}PV}$ *=.759,* $ICC_{PV\text{-}GF}$ *=.723*) but *LTS-LS* (*ICC =.569*). *LS-PV* furthermore presented good agreement between days except for *ROT* (*ICC =.396*). The error of α_{MAX} measurements (*Bias*) for *FLEX* was larger in the more caudal *Joints* (*LS-PV =18rad/s²* and *PV-GF =16rad/s²*), compared to more cranial *Joints* (*UTS-LTS =-6 rad/s²* and *LTS-LS =1rad/s²*). Though, more caudal *Joints* hereto presented larger Means of α_{MAX} in both measurements. (*Figure 1.5, Table A1.5*)

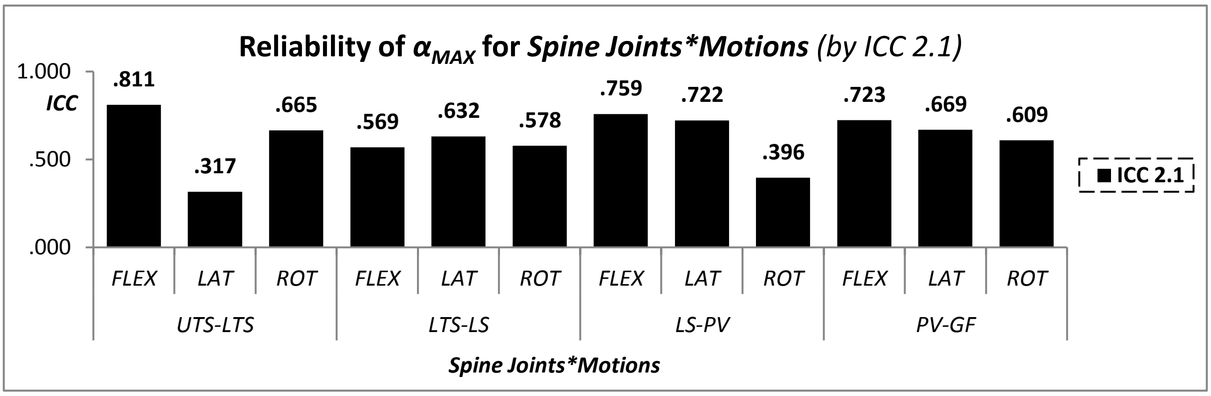


Figure 1.5: Between-day Reliability for α_{MAX} by *ICC 2.1* for specific *spine Joint*Motion* combinations

LANDING TYPES*HEIGHTS. Within landing *Types*: *UL* showed best agreement of α_{MAX} measurements between days (ICC_{UL60} =.738, ICC_{UL45} =.648, ICC_{UL30} =.675, ICC_{UL15} =.736). Though, measurement errors were generally larger in *UL* ($Bias_{UL60}$ =17rad/s^2, $Bias_{UL45}$ =28rad/s^2, $Bias_{UL30}$ =15rad/s^2, $Bias_{UL15}$ =7rad/s^2) compared to other landing *Types,* which were consistently as small as *0 to 5rad/s*^{2}. By far lowest agreement of α_{MAX} measurements between days was found at *BL60* (*ICC =.004*), *BL15* (*ICC =.285*), *BF15* (*ICC =.321*) and *BF30* (*ICC =.217*). (*Figure 1.6, Table A1.6*)

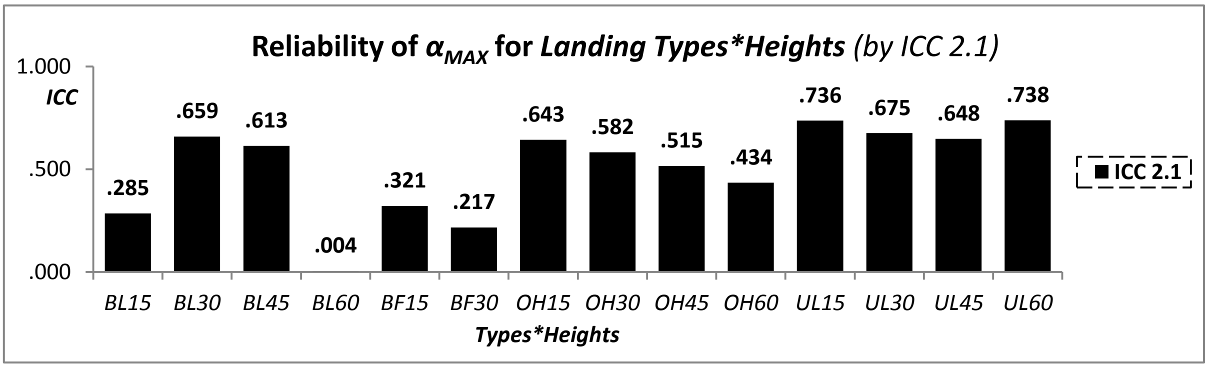


Figure 1.6: Between-day Reliability for α_{MAX} by *ICC 2.1* for specific landing *Type*Height* combinations

The, for reference conducted, reliability analysis on GRF_N indicated an overall high, superior reliability of vertical ground force measures (*M1: 2116 ±562N, M2: 2083 ±475N; Min: 569N, Max: 8332N; ICC: .901; TRV: 7%; Bias ± LoA: 151 ±401N*).

S1 - DISSEMINATION OF α_{MAX} ACROSS SPINE MOTIONS

α_{MAX} presented a general cranial magnitude decline along the spinal course in each motion plane. This was presented across and within particular landing *Types*Heights* for the average of the whole cohort, as for each individual subject; even though individual subjects' α_{MAX} magnitudes were largely divergent (*Figure 1.7*, *Table A1.8*).

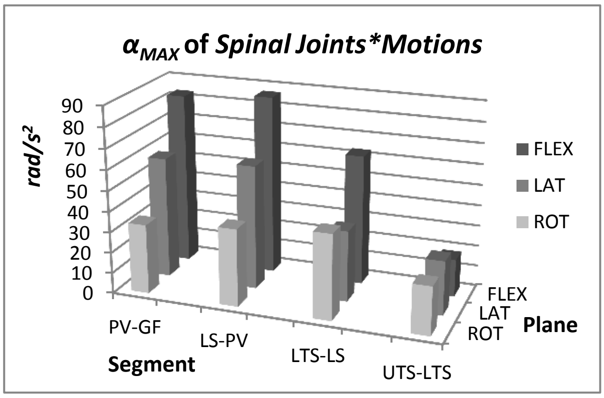


Figure 1.7: α_{MAX} of Spine *Joint*Motions*; averaged from all landing *Types*Heights*

Between spine *Motions, FLEX[LS-PV]* on the average of all landing *Types*Heights* showed significantly greatest α_{MAX} (*87 ±57rad/s²*) followed by *LAT[LS-PV]* (*60 ±35rad/s², p= .003*), followed by *ROT[LS-PV]* (*38 ±20rad/s², p= .000*). This rank order was consistently presented with statistical significance in each bipedal landing *Type*. Furthermore, across all landing *Types*Heights, α_{MAX}* in *FLEX[LS-PV]* was, with almost equivalent magnitudes to the sagittal accelerations of *PV-GF* (*83 ±46rad/s²*), larger compared to cranially following *Joints*: *FLEX[LTS-LS] =63 ±34rad/s²; FLEX[UTS-LTS] =18 ±10rad/s²*). This cranial decline was evenly found in each landing *Type*Height*, wherein only the differences between *FLEX[LS-PV]* and *FLEX[UTS-LTS]* were statistically significant. *UL* exceptionally presented highest α_{MAX} in *LAT[LS-PV]*, with equally highest lateral flexion acceleration of *[PV-GF]*, followed by *FLEX[LS-PV]* and *ROT[LS-PV]*. According to bilateral landings α_{MAX} in *UL* showed a substantial cranial decline in *FLEX* and *LAT*. (*Figure 1.7, Table A1.8*)

S1 - EMERGENCE OF $\alpha_{FLEX[LS-PV]}$ IN LANDING TYPES

The presentation of $\alpha_{FLEX[LS-PV]}$ in each landing *Type*Height* showed an apparent increase of $\alpha_{FLEX[LS-PV]}$ with increasing heights from *15* to *45cm*. Across landing *Types,* slightly highest $\alpha_{FLEX[LS-PV]}$ was found in *OH*, which was, depending on landing *Height*, followed by *BF* at *15* and *30cm*, and by *UL* at *45* and *60cm*. (*Figure 1.8*)

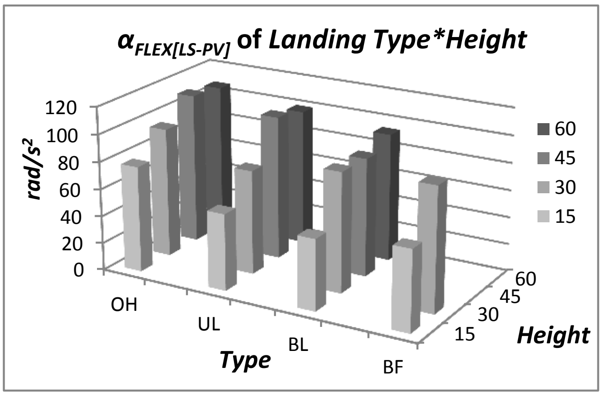


Figure 1.8: $\alpha_{FLEX[LS-PV]}$ of landing *Type*Heights*

GRF_N, used as ground reference load of each landing *Type*Height*, showed in contrast to $\alpha_{FLEX[LS\text{-}PV]}$ consistent increases in force magnitude with increasing *Heights* and a distinct order of landing *Types* accommodated with larger ground impact forces (*BL* < *OH* < *BF* < *UL*). Peak ground reaction forces were significantly greatest in *UL,* while differences between other landing *Types* were less substantial (*Figure 1.9, Table A1.9*). Despite these minor disagreements between GRF_N and $\alpha_{FLEX[LS\text{-}PV]}$ data, a sig. strong positive linear associations of GRF_N to α_{MAX} was exhibited overall: $r(1200)= .65$, $p= .000$ and within each *Joint* and *Motion*.

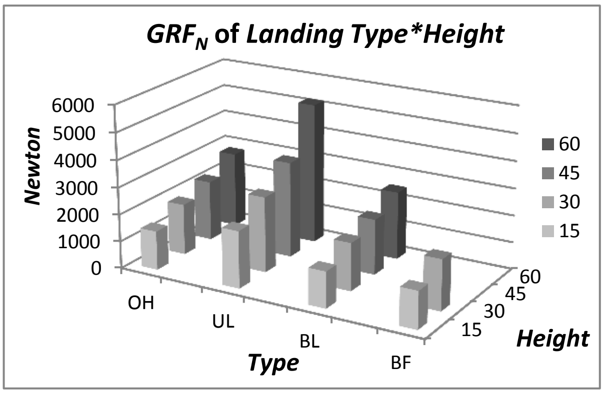


Figure 1.9: GRF_N of landing *Type*Heights*

S2 - Within-day Reliability of $\alpha_{FLEX[LS-PV]}$ within a Large Cohort and Homogeneous Groups

Upon an initial pre-evaluation for plausibility of the data from our second study, overall *5%* of measures were discarded. Hereof *94%* of total dropout measures were assigned to two subjects, who had to be reasonably excluded by: one female being not normal weight, based on the *BMI* assessment, creating outlier data on almost all assessed parameters, and one male subject presenting peak expressions of $\alpha_{FLEX[LS-PV]}$ implausibly and substantially earlier and larger than all other subjects, which was assumed to be founded on previously undetected assessment errors. (*Table A2.1*)

Cohort and Group Characteristics.

Mean age and anthropometrics of the analysed larger cohort (*n=43*) were *29.6 ±4.6yrs*; *1.72 ±0.09m*, *67.7 ±12.9kg*. Groups of *Familiarity* and *Gender* showed no substantial differences in age. Group differences in height and weight were marginal and non-significant between *Familiar* (1) and *Unfamiliar* (2) subjects (1_{WEIGHT}: *69.7 ±13.6kg*, 2_{WEIGHT}: *65.0 ±11.2kg*, $t(41)= 1.18$, $p= .246$) and (1_{HEIGHT}: *1.72 ±0.09m*, 2_{HEIGHT}: *1.73 ±0.10m*, $t(41)= -0.12$, $p= .907$). Though groups of *Gender* (*F, M*) showed substantial sig. differences in both anthropometric measures (F_{WEIGHT}: *57.9 ±5.7kg*, M_{WEIGHT}: *79.1 ±8.6kg*, $t(41)= -9.46$, $p= .000$) and (F_{HEIGHT}: *1.65 ±0.05m*, M_{HEIGHT}: *1.81 ±0.05m*, $t(41)= -10.40$, $p= .000$). (*Table A2.1*)

Sensitivity of $\alpha_{FLEX[LS-PV]}$.

The overall rate percentage of ground impact affiliated $\alpha_{FLEX[LS-PV]}$ [$t\text{-}IGC<t\text{-}\alpha_{FLEX[LS-PV]}(\%)$] was *82%*. *DJ* herein showed about *10%* lower rates compared to *DL* and *DS* overall and within groups. Landing *Unfamiliar* subjects showed across all landing *Types* *8%* to *24%* higher rates of $t\text{-}IGC<t\text{-}\alpha_{FLEX[LS-PV]}$ compared to landing *Familiar* ones. Even stronger differences were shown between groups of *Gender*, whereat *Females* (*F*) presented rates of impact affiliation to be *≈20%* lower compared to *Males* (*M*), with their strongest divergence in *DL* ($t\text{-}IGC<t\text{-}\alpha_{FLEX[LS-PV](F)} = 73\%$; $t\text{-}IGC<t\text{-}\alpha_{FLEX[LS-PV](M)} = 97\%$). (*Table 2.1*)

Table 2.1: Ground impact affiliation of $\alpha_{FLEX[LS-PV]}$ ($t\text{-}IGC<t\text{-}\alpha_{FLEX[LS-PV]}(\%)$) in cohort and groups of *Gender, Familiarity*

	Total	*DL*	*DJ*	*DS*
Cohort	82	84	74	84
Familiar (1)	77	79	64	80
Unfamiliar (2)	89	91	88	88
Females (F)	73	73	64	76
Males (M)	93	97	85	93

INTER- & INTRA-INDIVIDUAL VARIABILITY OF $\alpha_{FLEX[LS\text{-}PV]}$.

Overall inter-individual variability of $\alpha_{FLEX[LS\text{-}PV]}$ was large (CV_{AV} =52%). Across all landing *Types*, *Unfamiliar* subjects showed substantially lower inter-individual variability on $\alpha_{FLEX[LS\text{-}PV]}$ (CV_{AV} =37%) compared to *Familiar* ones (CV_{AV} =61%). Furthermore *Males* in total and in subgroups of *Familiarity* showed lower inter-individual variability on $\alpha_{FLEX[LS\text{-}PV]}$ compared to *Females* (*Table A2.2a,b*). Intra-individual variability (*CV ≈30%*) was lower than inter-individual assessments. Herein *Unfamiliar* subjects showed only marginally lower variability in $\alpha_{FLEX[LS\text{-}PV]}$ (*CV ≈24%*) compared to *Familiar* subjects (*CV ≈34%*), wherein highest difference between the two groups was found in *DJ* (*CV ≈26%, CV ≈40%*), (*Table A2.3a,b*). However ranges of $\alpha_{FLEX[LS\text{-}PV]}$*'s* intra-individual variability across subjects were very large (CV_{MIN} =9%, CV_{MAX} =96%).

Inter-individual variability of muscular *PRE*-activity was high for all abdominal muscles (CV_{AV} ≈49%) but lower for *Erector Spinae* (CV_{AV} =31%), (*Table A2.2a,b*). Muscle specific variability within subjects was by *≈10* to *30%* lower than between subjects (*Table A2.2b, Table A2.3b*)

S2 - Landing Type-, Landing Familiarity- and Gender Effects

Peak Lumbo-Pelvic Flexion Acceleration ($\alpha_{FLEX[LS\text{-}PV]}$).

Landing impact affiliated peak lumbo-pelvic flexion acceleration ($\alpha_{FLEX[LS\text{-}PV]}$) occurred on an overall average magnitude of *50 ±26rad/s²*. The range of $\alpha_{FLEX[LS\text{-}PV]}$ was given by a large dispersion across subjects (*MIN*: *9* to *MAX: 244rad/s²*), which was similarly found within each landing *Type* and in each group of *Familiarity* and *Gender* (*Figure A2.1*). *ANOVA* didn't show any sig. interaction effect for $\alpha_{FLEX[LS\text{-}PV]}$. A sig. main effect on $\alpha_{FLEX[LS\text{-}PV]}$ was presented for *Familiarity* $F(1, 39)= 5.93$, $p= .020$. This was, as shown by pairwise post-hoc comparisons, due to sig. smaller $\alpha_{FLEX[LS\text{-}PV]}$ in the *Familiar* group (*1*), consistently found in each landing *Type*: *DL* (*1: 43 ±30rad/s²; 2: 64 ±24rad/s², p= .017*), *DJ* (*1: 40 ±25rad/s²; 2: 56 ±21rad/s², p= .037*), *DS* (*1: 45 ±25rad/s²; 2: 60 ±21rad/s², p= .042*), (*Figure 2.2*, *Table 2.2*). Drop jumps furthermore appeared to reduce $\alpha_{FLEX[LS\text{-}PV]}$ in *Females* and *Unfamiliar* subjects, which was however not statistically significant.

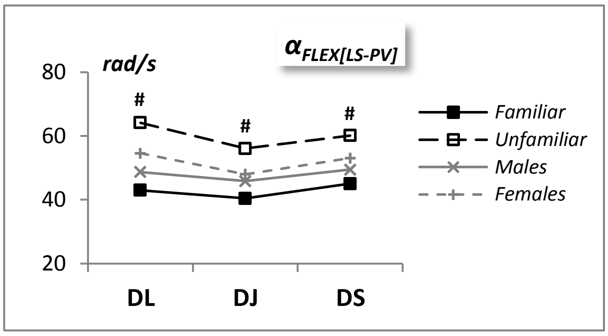


Figure 2.2: Group (*Familiarity, Gender*) and landing *Type* (*DL, DJ, DS*) effects on $\alpha_{FLEX[LS\text{-}PV]}$; *# sig. between subjects effect (p<0.05)*

Table 2.2: $\alpha_{FLEX[LS\text{-}PV]}$ (*Mean±SD*) for cohort and groups (*Familiarity, Gender*) within each, and on average of all landing *Types* (*DL, DJ, DS*)

	$\alpha_{FLEX[LS\text{-}PV]}$ (rad/s²)			
	DL	**DJ**	**DS**	**ALL**
Familiarity				
Familiar (1) *(n=25)*	**43** # ±30	**40** # ±25	**45** # ±25	**43** # ±26
Unfamiliar (2) *(n=18)*	**64** ±24	**56** ±21	**60** ±21	**60** ±22
Gender				
Female (F) *(n=23)*	**55** ±34	**48** ±27	**53** ±27	**52** ±29
Male (M) *(n=20)*	**49** ±23	**46** ±22	**50** ±22	**48** ±22
Cohort				
(n=43)	**52** ±29	**47** ±25	**51** ±24	**50** ±26

sig. different from respective compared group (*p<0.05*) = between subjects effect

TRUNK MUSCULAR ACTIVITY.

Trunk muscular *PRE*-activity showed highest overall *%MVC* values for Transversus Abdominis & Internus Obliquus ($[TrA\text{-}IO]_{PRE}$ =66 ±32%MVC) followed by Erector Spinae ($[ES]_{PRE}$ =47 ±15%MVC). Externus Obliquus and Rectus Abdominis instead presented substantially smaller cohort averages ($[EO]_{PRE}$ =15 ±8%MVC and $[RA]_{PRE}$ =10 ±4%MVC). Overall ranges (*MIN* to *MAX*) of muscular activity within the cohort were predominantly large for each muscle; though, margins of these ranges declined in the same sequence as muscular averages $[TrA\text{-}IO]_{PRE}$ (*11 to 194%MVC*); $[ES]_{PRE}$ (*23 to 105%MVC*); $[EO]_{PRE}$ (*4* to *43%MVC*); $[RA]_{PRE}$ (*4 to 20%MVC*).

ANOVA revealed sig. main effects on the repeated measure of landing *Type* for each muscle ($[TrA\text{-}IO]_{PRE}$: $F(2, 78)= 8.52$, $p= .000$; $[ES]_{PRE}$: $F(2, 78)= 29.3$, $p= .000$; $[EO]_{PRE}$: $F(2, 78)= 9.70$, $p= .000$; $[RA]_{PRE}$: $F(2, 78)= 7.83$, $p= .001$). Hereby each muscles' activity in *DJ* was significantly larger compared to *DL* and *DS*, except $[TrA\text{-}IO]_{PRE}$, which did not reach statistical significance between *DJ* and *DL*. Within groups of *Familiarity* and *Gender*, differences between *DJ* to *DL* and *DS* were overall consistent for $[ES]_{PRE}$, but less for abdominal muscles. For abdominal muscles *Males* and *Unfamiliar* subjects showed fewer statistically sig. differences between landing *Types* compared to *Females* and *Familiar* subjects.

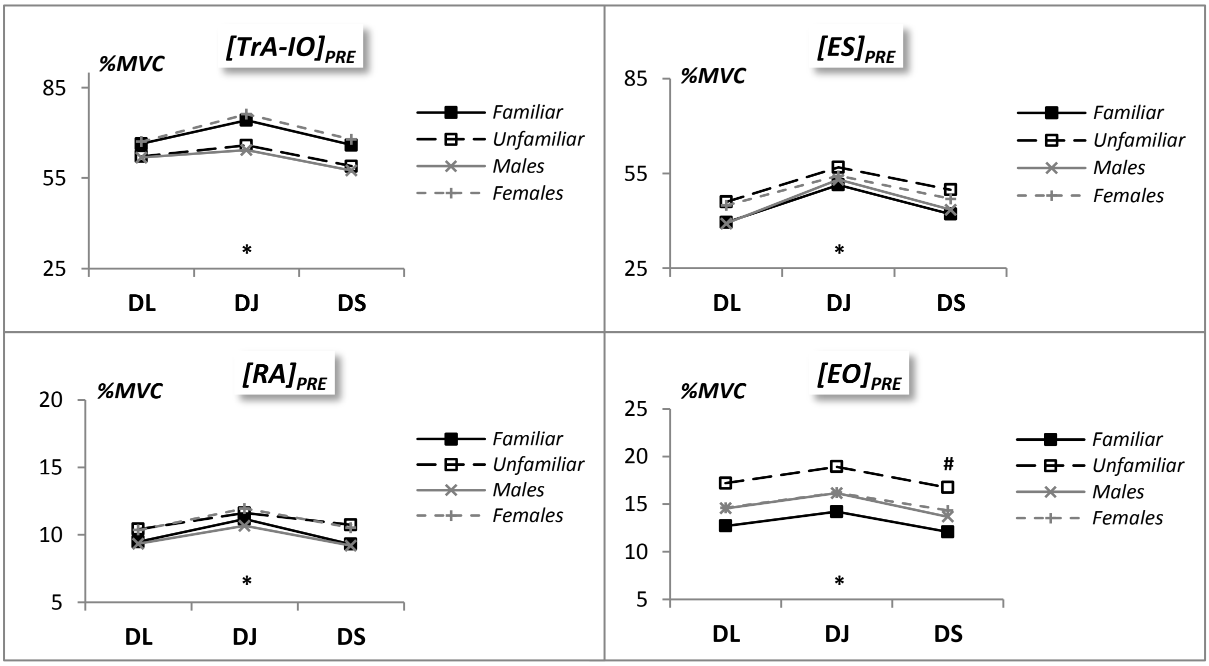


Figure 2.3: Group (*Familiarity, Gender*) and landing *Type* (*DL, DJ, DS*) effects on muscular *PRE*-activity: *[TrA-IO], [RA], [EO], [ES]* in *%MVC* ;

 # sig. between subjects effect (p<0.05)

** sig. within subjects effect (p<0.05)*

Females presented compared to *Males* overall higher activity of each muscle in each landing *Type*; with most remarkable difference of *[TrA-IO]*$_{PRE}$ in *DJ* (*F: 76 ±38%MVC*; *M: 64 ±31%MVC*). Though, no *Gender* difference was stat. sig. among any single muscle comparison. *[TrA-IO]*$_{PRE}$ activity was moreover higher in *Familiar* (*1*) compared to *Unfamiliar* subjects (*2*) in *DL* (*1: 66 ±29%MVC, 2: 62 ±28%MVC*), *DJ* (*1: 74 ±39%MVC, 2: 66 ±30%MVC*) and *DS* (*1: 66 ±35%MVC, 2: 59 ±25%MVC*). Furthermore, *Familiar* subjects showed high relative differences of *[EO]*$_{PRE}$ -activity compared to *Unfamiliar* ones, which reached stat. sig. at the pairwise comparison in *DS* (*1: 12 ±7%MVC, 2: 17 ±8%MVC, p= .044*). (*Figure 2.3*, *Table A2.4*)

Co-contraction Index for *[EO|ES]*$_{PRE}$ was presented by an overall average of *20 ±12%MVC* activity in a wide cohort range from *5 to 61%MVC*, which didn't vary remarkably between landing *Types*. No significant interaction effect for landing *Type*Familiarity*Gender* group matrix was revealed by *ANOVA* (*Figure 2.4*). However, *CCI[EO|ES]*$_{PRE}$ showed a sig. main effect for landing *Type F(2, 78)= 8.80, p= .001.* Hereto *post-hoc* pairwise comparison between landing *Types* showed minor but sig. differences between *DJ* (*22 ±12%MVC*) and *DS* (*19 ±11%MVC, p= .000*) for the whole cohort as for the comparative groups of *Gender* and *Familiarity* (*Table A2.5*).

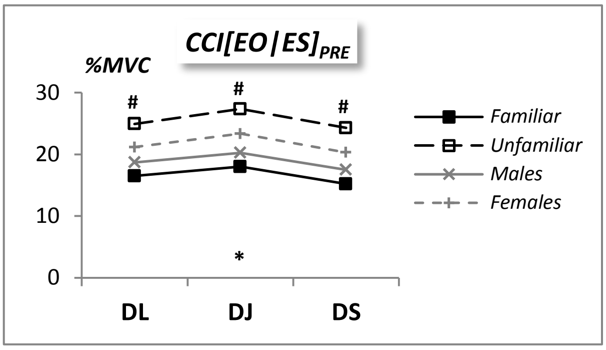


Figure 2.4: Group (*Familiarity, Gender*) and landing *Type* (*DL, DJ, DS*) effects on *CCI[EO|ES]*$_{PRE}$;

 # sig. between subjects effect (p<0.05)
** sig. within subjects effect (p<0.05)*

A significant main effect on *CCI[EO|ES]*$_{PRE}$ was shown for *Familiarity F(1, 39)= 6.99, p= .012*, revealing substantially lower co-contraction ratios of *Familiar* subjects (*1*) compared to *Unfamiliar* ones (*2*) in each landing *Type* (*DL*$_{1:}$ *17 ±9%MVC,* $_{2:}$ *25 ±14%MVC, p= .017; DJ*$_{1:}$ *18 ±9%MVC,* $_{2:}$ *27 ±15%MVC, p= .013; DS*$_{1}$*: 15 ±8%MVC,* $_{2:}$ *24 ±14%MVC, p= .008)*, (*Table A2.5*). In contrast *Gender* groups showed much lower non-sig. differences overall and in each landing *Type.*

Phase specific Deployment of Individual Trunk Muscles. Trunk muscles deployed the majority of their overall landing related contribution to the *PRE*-phase of landing, hence prior to the onset of $\alpha_{FLEX[LS\text{-}PV]}$. Herein, abdominal muscles contributed similar larger percentage of their *PRE*-phase shares (*[RA]*$_{PRE}$ *=86%*$_{LANDING}$, *[TrA-IO]*$_{PRE}$ *=81%*$_{LANDING}$, *[EO]*$_{PRE}$

$=78\%_{LANDING}$) compared to $[ES]_{PRE}$ ($69\%_{LANDING}$). Each muscle showed statistically significant higher contribution to *DROP* compared to *IMPACT* in each landing *Type*. Hereby, on the average of all landing *Types*, *[ES]* presented the least distinction between *DROP* ($39 \pm 10\%_{LANDING}$) and *IMPACT* ($30 \pm 7\%_{LANDING}$, $p= .000$). All abdominal muscles instead, deployed substantially larger shares of their overall activity to *DROP* compared to *IMPACT* (Figure 2.5, Table A2.6, Table A2.7).

By contrast, *%MVC*-normalised trunk muscle activities presented largely disparate activation amplitudes between muscles and between landing phases. Herein *%MVC* muscular activity of each muscle but *[RA]* was significantly lower during *DROP* compared to *IMPACT* (Figure 2.5, Table A2.8, Table A2.9). Muscular *%MVC* activity in *DJ*, was compared to other landing *Types* consistently higher for *[TrA-IO]*, *[ES]* and *[EO]* in both phases, which was accentuated in the *IMPACT* phase for *[TrA-IO]* and *[ES]*, (Table A2.8, Table A2.9).

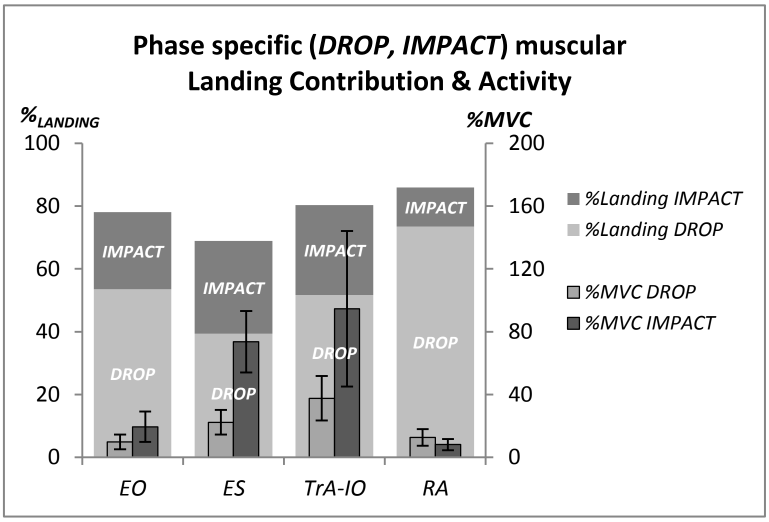


Figure 2.5: Phase (*DROP, IMPACT*) specific muscular contribution ($\%_{LANDING}$) and activity (*%MVC*) of *[EO], [ES], [TrA-IO], [RA]*

Phase specific muscular contribution ($\%_{LANDING}$) were generally identical for each group of *Familiarity* and *Gender*, whereas relative group differences didn't exceed *5%* (Figure A2.2, Figure A2.3, Table A2.6, Table A2.7). Though, *Familiar* subjects compared to *Unfamiliar* ones showed stat. sig. lower *%MVC* activity of *[EO]* during *DROP* and *IMPACT* overall (Figure 2.6) and in each landing *Type* (Table A2.8). Moreover in *DS; %MVC*-activity of $[ES]_{DROP}$ was stat. sig. lower in *Familiar* (*1: 18 ±6%MVC*) compared to *Unfamiliar* subjects (*2: 22 ±8%MVC, t(41)= -2.223, p= .032)*. *Females* compared to *Males* presented consistently higher muscular *%MVC* activity in each muscle at both phases, with highest relative differences in $[ES]_{DROP}$. These differences were found to be stat. sig. overall (Figure 2.7) and in each landing *Type* (Table A2.9).

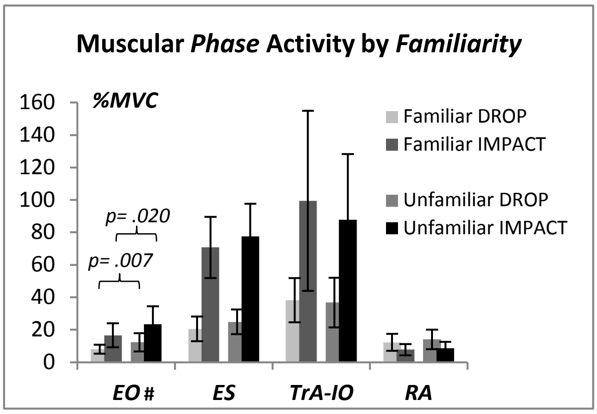


Figure 2.6: Muscular activity (*%MVC*) contrasts (*Mean±SD*) for *Familiarity* and landing *Phase* across all landing *Types*; **#** sig. group difference (*p<0.05*)

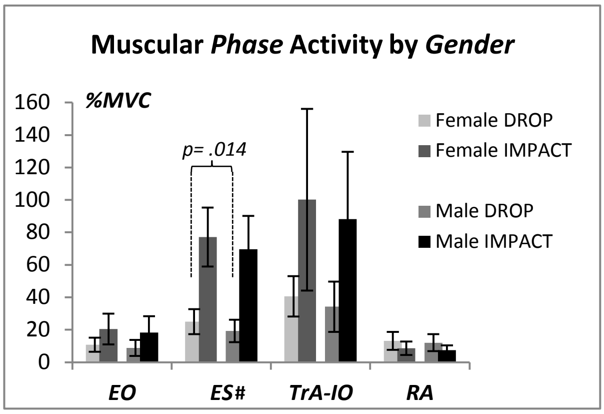


Figure 2.7: Muscular activity (*%MVC*) contrasts (*Mean±SD*) for *Gender* and landing *Phase* across all landing *Types*; **#** sig. group difference (*p<0.05*)

Trunk Muscular Activity Associations to Peak Lumbo-Pelvic Flexion Accelerations.

Associations of *%MVC* normalised muscular activity to $\alpha_{FLEX[LS\text{-}PV]}$ were weak and diverse across landing *Type*s. Strongest sig. pos. moderate associations herein were found between $[EO]_{PRE}$ and $\alpha_{FLEX[LS\text{-}PV]}$ in *DL*: r_S= *.39, p= .009* and in *DS*: r_S= *.31*, *p= .043*. *Males* hereto showed accentuated significant pos. moderate to strong associations between $[EO]_{PRE}$ and $\alpha_{FLEX[LS\text{-}PV]}$ across all landing *Types* (DL_M: r_S= *.68*, *p= .001*; DS_M: r_S= *.65*, *p= .002*) with slightly lower strength in (DJ_M: r_S= *.45, p= .045*). This was in large contrast to *Females*, in which such associations were not presented (*Table A2.10*). These associations for *[EO]* overall and within groups of gender were moreover equally found on similar strength in both *DROP* and *IMPACT* phase (*Table A2.11*).

Almost equivalent associations were revealed between *CCI[EO|ES]* and $\alpha_{FLEX[LS\text{-}PV]}$, which were weak pos. for $CCI[EO|ES]_{PRE}$ across all landing *Types*, and merely significant in *DL*: r_S=*.31, p= .042*. Remarkably in the subgroup of *Males*, these associations appeared as sig. moderate in *DJ* (*M:* r_S= *.53*, *p= .016*) and strong in *DL* (*M:* r_S= *.72*, *p= .000*) and *DS* (*M:* r_S= *.71*, *p= .000*). These associations in *Males* presented strongest substantiation in the *IMPACT* phase of landings, whilst appearing as continuously moderate in the *DROP* phase of each landing (*Table A2.11*). Females in contrast didn't present any meaningful association between *CCI[EO|ES]* and $\alpha_{FLEX[LS\text{-}PV]}$.

Within groups of *Familiarity,* associations between muscular activity and $\alpha_{FLEX[LS\text{-}PV]}$ were overall non-significant and weak. Though, one exceptional sig. moderately positive association between $[RA]_{PRE}$ to $\alpha_{FLEX[LS\text{-}PV]}$ in *DL* was found in *Unfamiliar* subjects (r_S *= .57, p= .013*), (*Table A2.10*). This association in *Unfamiliar* subjects was frequently found as moderate to strong during *IMPACT* in each landing *Type* ($DL_{\underline{2}}$: $r_S = .64$, $p = .004$, $DJ_{\underline{2}}$: $r_S = .65$, $p = .003$, $DS_{\underline{2}}$: $r_S = .43$, $p = .073$), (*Table A2.11*).

No phase specific muscular contribution ($\%_{LANDING}$) showed a significant association to $\alpha_{FLEX[LS\text{-}PV]}$, despite $[RA]_{DROP}$ which was negatively associated to $\alpha_{FLEX[LS\text{-}PV]}$ in *Females* during *DJ* (F: $r_S = .42$, $p = .044$), (*Table A2.12*).

DISCUSSION

OBJECTIVITY OF PEAK SPINE SEGMENTAL ANGULAR ACCELERATIONS (α_{MAX})

SENSITIVITY.

The assessment of spine peak segmental angular accelerations [α_{MAX}] appears to be a conditionally valid surrogate measure for striking spinal bending load in drop landings. Though, major regard has to be payed to α_{MAX}'s sensitivity of depicting the most energetic bending load attributed to the landing impact. Hereto our analysis showed that peak spine segmental angular accelerations occur only in *64 to 96%* of landings in succession to the landings ground impact; which appears to be moreover depending on the landing and performer characteristics. Our analysis hereto presented a strong positive relationship of α_{MAX}'s landing impact affiliation to α_{MAX}'s magnitude, indicating that α_{MAX} rather constitutes a valid measure for impact related peak spinal bending loads when severe landing impacts emerge. According to literature, our investigation showed that larger landing heights (*Zhang et al., 2000; 2008*) and body weight (*Sell et al., 2010*) extrinsically increase impact magnitudes by larger accumulation of gravitational potential energy, whilst performers' intrinsic landing techniques can antagonistically alleviate impact severities (*Santello, 2005; Pappas et al., 2012; Bruton et al., 2013*). Despite these findings, α_{MAX} in more cranial spinal joints seem, at lower magnitudes, less frequently affiliated to the ground impact. However, even in kinetically most vigorous landings and less skilled performers, as in most caudal spine joints, few measured peak angular accelerations appeared before the landing's ground impact. Deductively, the individuals measured in our landing trials did not hold their spines to a rigid posture before ground impact, hereby occasionally presenting more impulsive motion during the landings' dropping phase than successive to the ground impact. Perhaps our test subjects did not consistently manage to execute their minimal jump take-off as smoothly as desired for our aims. Hence, they hereby likely implemented distorting trunk posture adjustments during their dropping phase, which were rather attributed to their jump. This possible explanation introduces the question: Why these possibly hereof evoked angular accelerations were larger than those triggered by the landing impact? Hereto it is arguable that trials, wherein premature α_{MAX} were detected, rather presented considerable low angular acceleration magnitudes in the impact phase instead of substantially high accelerations in the dropping phase. This assumption was supported by the presented positive association between α_{MAX}*'s* magnitude to its impact affiliation and moreover conceivable by common very low magnitudes within the ranges in which α_{MAX} occurred.

WITHIN-DAY RELIABILITY (INTRA- & INTER-INDIVIDUAL).

One of the greatly concerned characteristics of peak spine segmental angular accelerations, measured during landings, was the presumed substantial intra- and inter-individual variability of this parameter. This presumption was confirmed by large ranges of intra-individual α_{MAX} magnitudes (*Min: 1, Max: 936rad/s²*) in our first, as repeatedly for $\alpha_{FLEX[LS\text{-}PV]}$ (*Min: 9.1, Max: 244.3rad/s²*) in our second study. Intra-individual variability of all assessed parameters was in both cohorts and in each distinguished group smaller than inter-individual variability, indicating overall more resemblance in landing pattern within individuals than within specific groups of individuals or landing motions. This finding generally accords to previous literature presentations for lower limb performances (*James et al., 2000; Santello. 2005; Nordin et al., 2016*) and to the findings of spine segmental motion variabilities in other impact perturbations (*Mueller et al., 2017*). Though, cohort averaged intra-individual variability of spine segmental α_{MAX} and $\alpha_{FLEX[LS\text{-}PV]}$ magnitudes (*CV =36%, CV =30%*) in our studies was yet significantly larger compared to previous literature findings on peak kinetic and peak angular kinematic parameters of the lower extremity joints (*Ford et al., 2007; Milner et al., 2011; Malfait et al., 2014; Alenezi et al., 2014*). Most likely, the in our studies substantial variability of α_{MAX} and $\alpha_{FLEX[LS\text{-}PV]}$ is assumed to be largely ascribed to inherently variable landing techniques of individuals (*James et al., 2000; Nordin et al., 2016*). Such variability within landing executions has been shown to be functionally natural. Adjustments in landing technique can significantly alter impact dissipation patterns (*Recknagel & Witte, 1996; DeVita & Skelly, 1992; Zhang et al., 2000; Santello. 2005*). Previous research has suggested an existence of optimal levels of intra-individual movement variability in human movement systems (*Davids et al., 2003*). Unfavourable deviations from that optimal movement variability emerge, when discrepancies between task demands and performer's skill exist (*James et al., 2000; Davids et al., 2003; Nordin et al., 2016*). *Nordin et al.* (*2016*) hereto verified that increased landing task demands (e.g. landing height increases) lead to a narrowing in the lower extremity joint degrees of freedom, accomplished by increased synergistic muscle activity and joint stiffness. Such however, would diminish impact dissipation at the affected joints (*Nordin et al., 2016*), and convey larger impacts to the spine. *Davids et al.* (*2003*) hereto moreover postulated that *"whilst unskilled performers tend to rigidly fix degrees of freedom", "skilled performers can freeze or unfreeze the degrees of freedom in a chain of movement as the prevailing task constraints demand"* (*Davids et al., 2003, p. 248*). Spine research in the same light has supposed such nonlinear dynamic aspects of system behaviour (*Reeves et al., 2007; Stergiou & Decker, 2011*). Nonlinear spinal sensory-motor behaviour facilitates a less confined dynamic compensation of external perturbations, thus enhancing spinal stability (*Reeves et al., 2007; Granata & England, 2007*). Consequently, variability of peak spine segmental angular accelerations must be perceived in context of a global synergistic interplay of multiple upper and lower body segments in a functional muscle-joint composition, conjointly pursuing the dissipation of the landing's ground impact (*Lees, 1981; Santello, 2005*). Individual's skill hereby capacitates the performer for either functionally utilising task affected joints for impact dissipation (*Lees, 1981; Davids et al., 2003; Santello, 2005*) or for preferably stiffening the same joints if potentially posed at risk to overload from the bending impact (*Davids et al., 2003; Nordin et al., 2016*). Based on our recruitment strategies, it is likely that a substantial number of individuals in both our studies have consistently or repetitively faced task demands exceeding their individual landing skill. In support of that perception, our analysis showed lower average intra-individual variability of $\alpha_{FLEX[LS\text{-}PV]}$ in the group of landing unfamiliar individuals accommodated by significantly increased levels of trunk muscular activity; indicating their less skilled approach of rather stiffening than functionally utilising the spinal joints. Notably, this approach appears to be impractical, as it contradictorily led to on average larger experienced $\alpha_{FLEX[LS\text{-}PV]}$ in this group.

Inter-individual variability of $\alpha_{FLEX[LS\text{-}PV]}$ was moreover found to be larger in females compared to males, indicating higher overall homogeneity of males' spinal stability. Previous studies have adjudged males with ordinarily superior landing skill, emerging from greater accumulated motor experiences with spinal impacts during childhood (*Bruton et al., 2013*). Contradictory to the hereof derived hypothetical conception of larger homogeneity in superiorly skilled cohorts, such as males, largest inter-individual variability of $\alpha_{FLEX[LS\text{-}PV]}$ was presented within the landing familiar group. This disconcerting finding is most probably ascribed to a debatable recruitment strategy of our study, which might have assembled a quite skill heterogeneous group of landing familiar individuals (*see chapter Limitations*).

All in all, indications by our data meet the presumptions about a substantial variability of spinal bending accelerations elicited by landing impacts. However an optimal level of intra-individual variability cannot be extracted from our results, due to the assumption that none of the groups in our study proffered an 'optimal skill level' matching the contrived task demands in our studies. Though, with regard to the spines supposed active integration in a holistic dynamic neuro-muscular dissipation of induced landing impacts, large variability of peak spine segmental angular accelerations might be physiological and therefore an inevitable part of spine segmental load evaluations.

Between-Day Reliability.

According to the high within-day variability *Test-Retest Variability* showed consistently high values above *24%*. Though, the between-day correlation analysis (*ICC 2.1*) revealed "moderate" to "good" agreements of α_{MAX} for most modelled spinal joints and motions as for most landing types. Systematic errors of measurement were found to be largest in most spinal flexion motions and in unilateral landings, which were reasonably explained by concomitantly highest α_{MAX} magnitudes. Ordinary bilateral landings instead showed overall lowest systematic errors of $0 rad/s^2$ to $3 rad/s^2$ with wide limits of agreement $\approx 30 rad/s^2$, which accorded to α_{MAX}'s large variability. The hereby presented between-day reliability of peak spine segmental angular accelerations was thus appraised as overall slightly poorer compared to between-day reliabilities previously reported for lower extremity peak joint angles and moments in various drop landings (*Ford et al., 2007; Milner et al., 2011; Malfait et al., 2014; Alezenzi et al., 2014*). Denotable, our analysis evaluated between-day reliabilities of spinal joints and motions aggregated across all administered landing types, as between-day reliabilities of particular landing types across all spinal joint motions. Within this approach, our reliability presentations are believed to be derated by particular landing types (e.g. *BF, OH*) and spinal motions (e.g. *ROT, LAT*), which recurrently showed poor reliability. While administered blindfolded and overhead-catch landings have uniquely been investigated by our study and can thus not be compared to other studies; reliability of transversal and coronal peak joint angles and moments of lower limb joints has been shown to be poorer than those in sagittal motions (*Ford et al., 2007; Milner et al., 2011; Alenezi et al., 2014*). This recurrent finding might be ascribed to smaller motion ranges expended in transversal and coronal motion planes during vertical landings (*Ford et al., 2007; Milner et al., 2011; Alenezi et al., 2014*). A by these previous authors frequently led discussion on potentially lower inter-session than intra-session reliability of lower extremity 3D-motion analysis was referred to possible errors in repeated marker placement. Such could alter modelled segments and joint centres, thus affecting angular motion captures (*Kadaba et al., 1989*). *Malfait et al.* (*2014*) reasonably trivialised that error potential in repeated marker placements, when functional joint centres are delineated by palpatable bony structures. By contrast, trunk marker placement for depiction of spine segments is a more inexplicit task for an investigator. Herein the less prominent vertebra structures are more difficult to identify by manual palpation, particularly in the upper regions of the spine, where spinious processi are barely detectable in many individuals. Actual positions of the transverse processi along the whole spine are furthermore entirely obscured and are in current literature modelled at a fixed distance on an imaginary superficial horizontal line (*Preuss & Popovich, 2010; Mueller et al., 2016a, b*). Consequently, these position markers are thereby situated on potentially moving soft tissue (*Mahallati et al., 2016*). Inter-segmental spinal angles have shown to be sensitive to errors due to such trunk marker placement, causing most concise errors in coronal and transverse plane motions (*Rouhani et al., 2015*). This biasing effect may in turn furthermore explain the in our study universally presented superior reliability in peak sagittal flexion accelerations within- and between-individuals as between days.

Larger reliability of α_{MAX} in sagittal flexion might be moreover attributed to concomitantly higher α_{MAX} magnitudes and the thereof augmented likeliness of being attributed to the landing's ground impact. In concordance with that, higher α_{MAX} magnitudes from elevated ground impacts in unilateral landings were accompanied by herein found overall higher between-day reliability. Higher reliability has accordingly been previously shown for larger joint moments at the lower limbs and for more impactful unilateral landings (*Alenezi et al., 2014; Ford et al., 2007; Milner et al., 2011*). Thus conclusively, peak spine segmental bending accelerations appear to provide a credibly objective parameter for ground impact induced spinal bending load evaluations in vigorous drop landings.

Appraisal of Peak Spine Segmental Angular Accelerations

Our observations registered traceable, most substantial spine segmental sagittal and lateral flexion accelerations dispersed across the whole spinal course. These presentations reveal the onsets of dispensed bending loads at the spine, moreover suggesting the spines participation in the dissipation of landing impacts. These findings are in agreement with previous research suppositions (*Recknagel & Witte, 1996; Santello, 2005; Kulas et al., 2006; Iida et al., 2011; Popovic & Kulig, 2012*). According to previous findings on lower extremity joints (*DeVita & Skelly, 1992; Zhang et al., 2000; Blackburn & Padua, 2008*), where predominant damping exigencies during bilateral vertical drop landings were described in the sagittal plane, highest peak segmental accelerations were continuously found in flexion along the spine. This is furthermore comprehensible by the prevailing impacting forces in that motion plane. Exceptionally in unilateral landings from larger heights (*45cm, 60cm*), greatest spinal bending accelerations emerged in lateral flexion. These, reasonably occurred in response to a presented sudden vigorous pelvic lateral tilt, caused by the missing area of support on the limb deprived side of the body. Such kinematic trunk response to unilateral landings has previously been shown by angular excursion and velocity measures of the lumbo-pelvic region (*Popovic & Kulig, 2012*).

Within a small heterogeneous cohort of 17 healthy individuals, recruited in our first study, overall highest peak flexion accelerations occurred at the lumbar spine. This finding indicates that active spinal impact load dissipation is predominantly implemented at this location, which is likely based on its larger structural compliance compared to the thoracic spine. The lumbar spine's greater skeletal compliance is assumingly met with employed muscular stiffness, bracing the lower trunk (*Cresswell et al. 1994; Hodges et al., 1996; Radebold et al., 2000; Lee et al., 2006; Vera-Garcia et al. 2007; Haddas et al., 2016,a,b*). Though, due to the muscles' inherent elastic properties, muscular stiffness, compared to skeletal and ligamentous joint formations, presumptively provides lower rigidity against large impact perturbations. In this light, previous literature has described spinal 'spring-like' damping properties (*Recknagel & Witte, 1996*), which have been showcased to depend on the degree of active stiffness (*Lees, 1981*). *Recknagel & Witte* (*1996*) hereto even supposed that, based on a 'spring-model', the large compliance of the lumbo-pelvic spine sector, may promote spondylolisthesis at this location provoked by high impact landings.

Unfortunately, a biomechanical or clinical appraisal of our findings is confined, since none of the consultable literature, addressing landing induced spinal loads (*Recknagel & Witte, 1996; Santello, 2005; Kulas et al., 2006; Iida et al., 2011; Popovic & Kulig, 2012, Ng et al., 2006; Zhang et al., 2008; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*), precedently determined spinal bending accelerations. Our repeated assessment of peak lumbo-pelvic flexion accelerations revealed average acceleration magnitudes of $87 rad/s^2$ in our first, and $50 rad/s^2$ in our second study. These averages were fairly similar to peak sagittal angular accelerations emerging at the lower extremity joints during diverse drop landings (ankle: $93.64\ rad/s^2$, knee: $117.57\ rad/s^2$, hip: $73.81\ rad/s^2$; *Siegmund et al., 2008*). Most precarious, however, were the extensive ranges, in which peak lumbo-pelvic flexion accelerations occurred in our assessments, evincing distinct individual average magnitudes of $244 rad/s^2$, and uniquely emerging magnitudes of even up to $506 rad/s^2$; notably these outputs were contained in the considered plausible dataset. In other research areas, peak trunk angular accelerations have been exposed as relevant load components within a cluster of trunk loading factors, which can increase the risk of occupationally related low back disorders (*Marras et al., 1993*). Peak angular trunk accelerations at high risk work places have herein been presented by, compared to our findings, low average (*92°/s*^{2} = *2rad/s*^{2}) and peak expressions (*514°/s*^{2} = *9rad/s*^{2}). This large divergence between our revealed measures compared to *Marras et al.*'s findings (*1993*) indicates a risk for acute overload to the lower trunk's dorsal soft tissues, which are assumed to be vigorously challenged by the evinced large landing induced flexion accelerations. This finding must be moreover regarded in the context of commonly presented high frequencies of landing executions by professional or recreational jumpers within single exercise sessions and across an overall active involvement in jumping sports throughout lifetime. However, an interpretation of supposed actual lumbar bending load needs to be drawn cautiously, when merely appraising peak spinal segmental flexion accelerations measured. As preliminarily acknowledged, sole segmental angular acceleration measures do not provide actual load dimensions. Though, valuable insights into the load onset dispersion across the spine and within variations of landing demands and performers' executions were revealed from our assessments.

Factors Contributing to Peak Lumbo-Pelvic Flexion Accelerations

Previous landing literature has well documented that landing impact severity is largely influenced by landing heights (*McNitt-Gray, 1993; Zhang et al., 2000, 2008*) and performers' weight (*Caster & Bates, 1995; Kulas et al., 2008, 2010; Janssen et al., 2012*), which account for the overall gravitational potential energy of any landing. Hereto our study revealed that single cases of obese individuals, incorporated in our first cohort, presented overall substantially larger peak lumbo-pelvic flexion accelerations; notably these occurred without concomitantly protruding peak ground reaction forces, which was unexpected. It hereby appears that augmented upper body mass might evoke enhanced spinal bending loads, by more than intrinsically elevated gravitational potential energy. Overweight has been previously shown to constrain landing skill and increase lower limb joint moments (*Minetti, 1998; McMillan et al., 2010*). Though, motor control of augmented body mass might pose an expansion of task demands, which might interfere with proper landing impact dissipation (*Santello, 2005; Sell et al., 2010, Nordin et al., 2016*). Thus intrinsic overweight apparently presents a multifactorial influence to peak lumbo-pelvic bending accelerations and loads. It hereby moreover appears that spinal impact load must be critically appraised by more than just by the landing's gravitational potential energy, accumulated by landing height and performers weight.

Effects of Landing Characteristics.

Our study exposed peak lumbo-pelvic flexion accelerations to emerge in non-linear dependence of landing heights; though sundry landing types appear to inherently engender altered $\alpha_{FLEX[LS\text{-}PV]}$. Herein, unilateral landings caused substantially larger bending accelerations across the whole spinal course along with highest ground reaction forces, which was reasonably explained by the lower area of support during landing and the herein deducted lower extremity muscular work, thus leading to larger and asymmetrically operating impacts at the pelvis. Moreover, most markedly, landings performed under obstructed focus to the landing execution (blindfolded, ball catching in airtime) provoked significantly increased peak lumbo-pelvic flexion accelerations. Landings under obstructed focus have found only rare attention in previous research; though, unanticipated vertical drop's, experimentally applied by *Recknagel & Witte* (*1996*), have been supposed to engender severe peak lumbo-pelvic accelerations, even from lowest dropping heights. Hereto *Santello et al.* (*2001, 2005*) suggested: that in absence of retrievable sensory-motor memories, visual information about landing heights are necessary for an environmental mapping, upon which a global feed-forward landing motor control is arranged to successfully dissipate landing impacts.

Landings entailing an immediate follow-up task (drop jump), by contrast, appear to increase spinal stiffness and hence engender lower peak lumbo-pelvic accelerations. Drop jump performances have been previously shown to increase joint stiffness at the lower limbs and hence reduce bending load components to the lower limb joints (*Ambegaonkar et al., 2011; Prieske et al., 2013, 2015a; Hackney et al., 2016*). Such apparent joint protective effect was argued with the performance attributed implementation of stiffness elevation, in order to reactively transpose the most feasible share of kinetic energy from dropping into an imminent forceful vertical push off (*Ambegaonkar et al., 2011*). Thus, due to the trunk's assumed essential contribution to this integral task implementation (*Santello, 2005; Iida et al., 2011; Prieske et al., 2013; Haddas et al., 2016*), a surmised accordingly elevated spinal stiffness, can be accredited for a protective effect against peak lumbo-pelvic flexion accelerations; even though these effects were not statistically significant and rather attributed to individuals with alleged poorer landing skill (landing unfamiliar and female individuals).

Landing under extrinsically delayed decision for an instant follow up jump (*DS*), did not affect peak lumbo-pelvic flexion accelerations. This presentation in our study, originated from the individuals' task realizations, which, as witnessed by the observer's visual inspection, failed the request of instantly jumping off, when such was required. Most individuals hereby instead implemented drop landings with delayed take-offs. Similar findings were previously reported on decision making landings (*Leukel et al. 2012; Mache et al., 2013*). Both these research groups coherently revealed that lower limb kinematic and musculo-tendinous performances are in case of follow-up task uncertainty rather by default conducted in favour of a mere landing execution. Hereof it can be deduced that integral impact dissipation might be the subconsciously favoured task implementation in landings, possibly to more effectively control and confine otherwise large compression loads (*Recknagel & Witte, 1996; Zhang et al., 2008; Panther & Bradshaw, 2013*).

EFFECTS OF TRUNK MUSCULAR PRE-ACTIVITY.

Pre-employed dorso-ventral co-contraction index ratios in our study showed, significant moderate associations to peak lumbo-pelvic flexion accelerations. Unexpectedly, these associations were positive and becoming stronger in the *IMPACT* phase of landing. This unexpected presentation was furthermore reiterated by group-specific larger co-contraction ratios at *IMPACT* and concomitantly increased $\alpha_{FLEX[LS\text{-}PV]}$. Comprehension of the calculated co-contraction index ratio, referred to *Lewek et al.* (*2004*), expresses that a presented larger co-contraction ratio between Erector Spinae and Externus Obliquus signifies a magnitude convergence of these selected dorsal and ventral muscular activities. Externus Obliquus in our observations acted on significantly lower percent maximal activity (*15%MVC*) than Erector Spinae activity (*47%MVC*). Our analysis moreover revealed that small activity elevations of Externus Obliquus particularly at *IMPACT*, occurred alongside significantly increased $\alpha_{FLEX[LS\text{-}PV]}$. Furthermore, Externus Obliquus activity showed overall positive significant associations to $\alpha_{FLEX[LS\text{-}PV]}$. Hence, an accentuation of Externus Obliquus activity would apparently disrupt spinal co-contraction, hereby promoting larger lumbo-pelvic flexion accelerations.

Our consultation of Erector Spinae and Externus Obliquus was motivated by previous findings, showing largest co-activity modulations of these muscles upon anteriorly directed sudden trunk perturbations (*Lavender & Marras, 1995; Krajcarski et al., 1999; Stokes et al., 2000*) and their preceding consideration in landing literature (*Iida et al., 2011; Popovic & Kulig, 2012*). *Popovic & Kulig* (*2012*) calculated, equally to our study, a co-contraction index ratio between Erector Spinae and Externus Obliquus during drop landings. Their revealed ratio between these two muscles differed from ours by lower Erector Spinae activity of *≈25%MVC*, which was likely based on differences in the *EMG* normalisation method (*see* *chapter Limitations*). Unfortunately, *Popovic & Kulig* (*2012*) merely consulted their calculated co-contraction index ratio in a single timeframe, rather to substantiate their between-group differences. In contrast, our phase differentiated activity observation revealed that Externus Obliquus activity demonstrated a remarkably smaller (*1.9*-fold) increase of activity from *DROP* (*10%MVC*) to *IMPACT* (*19%MVC*) compared to a *3.4*-fold increase of Erector Spinae from *DROP* (*22%MVC*) to *IMPACT* (*74%MVC*). Such divergent increase of dorsal and ventral muscular activity indicates that *CCI[EO|ES]* should not be consulted solitarily for an interpretation of spine stabilising muscular co-activations. Notably, Erector Spinae's large activity increase from *DROP* to *IMPACT* was exceptional among all assessed trunk muscular representatives. In view of this presentation, *Iida et al.* (*2011*) suggested that Erector Spinae predominantly pursues its assigned role of flexion attenuation in the *IMPACT* phase of landing and rather secondarily contributes to spinal stability. Contradictorily, elevated back muscle responses, shown by previous flexion loading trunk perturbation experiments (*Cresswell et al., 1994; Lavender & Marras, 1995; Krajcarski et al., 1999; Granata & Orishimo, 2001; Mawston et al., 2007*), have most consistently been endorsed for Erector Spinae's contribution to spine stabilising dorso-ventral co-activation. However, our interpretation of the substantial increase of Erector Spinae activity towards *IMPACT* concedes both previous comprehensions to be true; thus indicating that the antero-inferior bending impulse arising in landings must be sophisticatedly stabilised by additional trunk muscle representatives.

Interestingly the second largest increase of activity towards the *IMPACT* phase was seen in the, of all trunk muscles most active, Transversus Abdominis & Internus Obliquus; rising from (*38%MVC*) to (*95%MVC*). Transversus Abdominis & Internus Obliquus activity was disregarded in few previous landing studies (*Iida et al., 2011; Popovic & Kulig. 2012*). Though, *Kulas et al.* (*2006*) showed, according to our findings, substantially largest activity of Transversus Abdominis & Internus Obliquus compared to concomitantly recorded Externus Obliquus and Rectus Abdominis in drop landings. These, by our and *Kulas et al.*'s finding (*2006*), large locally detected activities lead to the assumption that Transversus Abdominis & Internus Obliquus have a considerable share to spinal stability in drop landing. In agreement with our findings, *Kulas et al.* (*2006*) emphasised a particular role of Transversus Abdominis & Internus Obliquus as particular local key dynamic stabilisers for the lumbo-pelvic region. This muscle group has been frequently turned out to predominately enhance lumbar-spinal and sacro-iliac joint stiffness, through its capability of effectively increasing intra-abdominal pressure (*Cresswell et al., 1994a, b; Hodges & Richardson,* *1997a; Krajcarski et al., 1999; Granata & Orishimo, 2001; Richardson et al., 2002; Mawston et al., 2007*). Both, Transversus Abdominis' and Internus Obliquus' functional role as lumbo-pelvic stabilisers can be well related to their anatomic insertions. Transversus Abdominis herein girdles the trunk like a waist harness from the Fascia Thoracolumbales to the Rectus sheath and can thus compress the lateral and abdominal wall in the transversal plane. Its abdominal descending fibres, proceeding fittingly along the Ligamentum Inguinale, can along with the at this location paralleled fibres of Internus Obliquus direct a posterior pelvic tilt, which leads to additionally abdominal compression in the sagittal plane. Denotable, Externus Obliquus equally possesses an abdominal compressing and pelvic tilting competence, due to its anatomical structure. Thus Externus Obliquus has also been frequently cited as collaborating in abdominal bracing (*Lavender & Marras, 1995; Krajcarski et al., 1999; Stokes et al., 2000; Haddas et al., 2016a, b*). Based on our results, it appears that the fellowship of each: Transversus Abdominis, Internus Obliquus and Externus Obliquus evoke a bracing compression of the abdominal wall particularly at the *IMPACT* phase of landing. As these muscles additionally induce a pelvic posterior tilt, thus creating trunk bending moment, they may concomitantly antagonise the trunk extending Erector Spinae activity. However, from our trials we cannot identify, if spinal robustness against the drop landing impact elicited trunk flexion perturbation is superiorly administered by abdominal bracing (*Kulas et al., 2006; Haddas et al., 2016a, b*) or dorso-ventral co-contraction (*Popovic & Kulig, 2012*). Instead, by the observation of altered muscular activities and $\alpha_{FLEX[LS\text{-}PV]}$ presentations in our investigation, interferences between the flexion attenuation remit of Erector Spinae, dorso-ventral co-contraction, and abdominal bracing might frequently occur in vertical drop landing performances.

With respect to previous literature suggestions about the conjunction of preparatory and reactive muscular control of landing impacts (*Santello, 2005; Kulas et al., 2006; Iida et al., 2011*), our analysis of $\%_{LANDING}$ related muscular contribution to landing phases showed totalled larger contribution to the *DROP* compared to the *IMPACT* phase by each trunk muscle recorded. Conversely however, phase-averaged *%MVC* activity of each but Rectus Abdominis muscle approximately doubled (Transversus Abdominis & Internus Obliquus; Erector Spinae) or tripled (Externus Obliquus) from *DROP* to *IMPACT*. Interpretation of these general findings has to be made with regard to the substantially longer *DROP* phase period of *≈300ms* compared to the much shorter *IMPACT* phase with *≈70ms*. Thus a relativized contemplation of our parameters with regard to the observed phase lengths signifies an overall lower activation of muscles in the *DROP* compared to the *IMPACT* phase. Merely Rectus Abdominis presented larger $\%_{LANDING}$ contribution and *%MVC* activity in the *DROP* phase compared to *IMPACT*. Though, this activity distribution was likely designated to a functional retraction of Rectus Abdominis towards the *IMPACT* phase of landing; in order to prevent undesired additional bending torque against Erector Spinae's trunk erection remit. Rectus Abdominis does not possess equivalent abdominal bracing competence as abdominal oblique and transverse muscles, and may have thus unpretentiously been retracted. The larger employment of Rectus Abdominis in the *DROP* phase instead might be referred to its strongest contribution to arrange a forward lean of the trunk after the jump-off (*Iida et al., 2011*), in order to promote more flexed and hence softer landing conduction (*Blackburn & Padua, 2008; Blackburn et al., 2009; Kulas et al., 2010; Iida et al., 2011*). Besides, Rectus Abdominis' overall activity of *10%MVC* was remarkably low compared to other muscles; Though Rectus Abdominis showed positive moderate associations to $\alpha_{FLEX[LS\text{-}PV]}$ at the *IMPACT* phase of landing, indicating a potential disruption of spinal robustness, when Rectus Abdominis is not retracted opportunely.

A very distinctive phase-activation feature was found for Erector Spinae; presenting compared to each abdominal muscle a significantly lower share of its overall landing contribution ($\%_{LANDING}$), along with relatively lower *%MVC* activity during *DROP*. This behaviour might be physiologically motivated by Erector Spinae's remit of attenuating trunk flexion at the *IMPACT* phase. Several previous experimental perturbation studies hereto showed that the magnitude of Erector Spinae activity towards flexion attenuation is depending on its inherent level of pre-activity (*Cresswell et al., 1994; Lavender & Marras, 1995*); whereat, for equal definite perturbation load magnitudes, lower pre-activity leads to larger perturbation responsiveness (*Krajcarski et al., 1999*). Elevated levels of muscular pre-activity and stiffness, in contrast, accompany an inhibiting effect on reflex elicitation, featured by prolonged reflex delays and diminished reflex magnitudes (*Stokes et al., 2000; Granata et al., 2004; Vera-Garcia et al., 2006*); which are assumed to ultimately result from lower stretch recognition by spindle units (*Granata et al., 2004*). Hence a trade-off between preparatory spinal stiffening and reflex utilisation has to be made when impact perturbations occur (*Granata & Marras, 2000 McGill et al., 2003*). Thus, the relatively lower activity of Erector Spinae, deployed in the *DROP* phase of our landing trials, may portray retained stiffness levels of paraspinal muscles, in order to react with improved flexion counteracting reflex contribution upon the imminent and considered indistinct landing impact. This proposition was furthermore supported by presentations of significant positive moderate associations between Erector Spinae activity and $\alpha_{FLEX[LS\text{-}PV]}$ in the *DROP* phase of landing.

In consideration of the mere short timeframe from touchdown to the onset of peak lumbo-pelvic flexion acceleration (*≈70ms*); spindle recognition based paraspinal reflexes are, due to their minimal delay of approximately *70ms* (*Granata et al., 2004*), considered as to slow for the counteraction of the early peak spinal bending impact. Hence, presumptively rather previously conditioned reflexes appear to be suitable as an efficient feed-back response for peak spinal bending impact diminution, due to their appearance in a predefined time course (*Santello, 2005*). Hereto it must be moreover argued, that for an accurate conditioned reflex response, prior anticipation of the actual impact magnitude and time is mandatory (*Santello et al., 2001; Mawston et al. 2007*). Such is not warranted under insufficient precursory familiarisation and skill acquisition. Moreover, particularly in unfamiliar landings, the unpredictability of the preceding crucial lower extremity damping performance, and its strong modifying effect on the initial ground impact magnitude (*Recknagel & Witte, 1996; Ng et al., 2006; Zhang et al., 2008; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*), might additionally obscure the anticipation of the trunk impact. Such was to some extend displayed in our previous study, showing onsets of substantially larger peak lumbo-pelvic flexion accelerations in landings where anticipation of the incoming impact was impeded in blindfolded (*BF*) and overhead task landings (*OH*). Hence in our second study: efficient conditioned trunk reflexes, alleviating peak lumbar bending accelerations, might have occurred only in individuals possessing a skillset of anticipatorily estimating the emerging impact characteristics. Such anticipatory proficiencies are assumed to be largely based on previously acquired sensory-motor memories from previous landing experiences.

EFFECTS OF PERFORMER SKILL FROM PREVIOUS FAMILIARITY AND GENDER.

Within our comparisons between several groups of individuals with presumptively different landing proficiencies, previous landing familiarity was revealed as the predominant determinant altering peak lumbo-pelvic flexion accelerations. Hereby landing familiar individuals presented on average significantly lower $\alpha_{FLEX[LS\text{-}PV]}$ magnitudes ($43rad/s^2$) compared to unfamiliar ones ($60rad/s^2$), when performing predictable and unpredictable drop landings and drop jumps. Concomitantly, landing unfamiliar individuals showed predominantly higher muscular activation with significantly elevated Externus Obliquus activity; whilst by contrast Transversus Abdominis & Internus Obliquus activity was substantially lower in this group. Landing unfamiliar individuals moreover displayed exclusive significant strong positive associations between $\alpha_{FLEX[LS\text{-}PV]}$ and Rectus Abdominis activity at *IMPACT,* indicating that this group presents a larger likelihood of failing to retract

Rectus Abdominis activity after adjusting the trunk's forward lean in succession to the take-off. That, along with the predominant elevation of trunk muscle activity indicates a rather crude spinal stiffening approach in landing unfamiliar individuals. Excessive spinal stiffening appears to be an unsuitable approach to counteract landing impacts, since it unfavourably allowed larger peak lumbo-pelvic bending accelerations; and is furthermore believed to increase spinal compression loads (*Van Dieën et al., 2003*). Assumingly, as previously argued with regard to abdominal muscles' trunk bending competences, the superelevation of Externus Obliquus and Rectus Abdominis activity might disruptively interfere with Erector Spinae's assigned bending counteraction. The contrastingly by landing familiar individuals presented elevated Transversus Abdominis & Internus Obliquus activity during the *IMPACT* phase of landing, alongside otherwise reduced muscular activity, appears to showcase a more effective muscular activation pattern, resulting in alleviated peak lumbo-pelvic flexion accelerations. Notably, due to its anatomical insertions, Transversus Abdominis is, when compared to abdominal oblique muscles, believed to generate less pronounced trunk sagittal flexion torque. Supposedly a deployed larger share of Transversus Abdominis might effectively raise intra-abdominal pressure, bracing the trunk without disturbing Erector Spinae's flexion counteraction. Previous studies have shown variously compound activity pattern of multiple abdominal muscles; including transverse, oblique, and longitudinally directed representatives, bracing the trunk against anteriorly directed impacting perturbations (*Cresswell et al., 1994a; Lavender & Marras, 1995; Krajcarski et al., 1999; Stokes et al., 2000; Kulas et al., 2006; Mawston et al., 2007*). Yet, subjects' prior familiarisation or experience with trunk perturbations was not consistently regarded by those authors. Few studies, however, have demonstrated that previous exposure to spine bending impacts leads to changes in the trunk's postural response (*Mawston et al., 2007; Pederson et al., 2004, 2007*). Thus, the altered trunk activation pattern seen in previous landing familiar individuals supposedly originated from these individuals' previous landing exposure and hereof acquired landing skill.

Based on previous literature indications our study aimed to verify suggested unequal landing proficiencies between males and females (*Decker et al., 2003; Kulas et al., 2006; Pappas et al., 2007; Kernozek et al., 2008; Weinhandl et al., 2010; Pappas & Carpes, 2012; Butler et al., 2013; Weltin et al., 2016*) and moreover supposed differences in spinal load (*Marras et al., 2002; Kulas et al., 2006*). Our findings didn't showcase significantly different peak lumbo-pelvic flexion accelerations in females ($52 rad/s^2$) compared to males ($48 rad/s^2$). Despite, females' muscular activation pattern showed compared to males' similar phase distribution of an overall marginally elevated muscular activity, in both phases of drop landing. Hereby our presentations diverge from *Kulas et al.*'s findings (*2006*), which showed females with significantly lower muscular pre-activation of Transversus Abdominis & Internus Obliquus in the drop phase of landing. *Kulas et al.* (*2006*) hereof concluded that males apparently employ a reinforced feed-forward abdominal muscle activation strategy, which females do not. This discrepancy in the showcased gender disparities is most likely owed by the deployed different *EMG* signal referencing methods. *Kulas et al.* (*2006*) hereby normalised each analysed muscle activity to signal amplitudes recorded from a single submaximal supine leg lifting task (*Dankaerts et al., 2004*). To our concern, this simplified method might not generate valid normalisation measures for the depiction of actual muscle activity proportions, since the hereby employed muscles are not presumed to have equal shares on holding the lifted legs. Moreover, with regards to dissimilar gravitational forces of gender disparate lower limb masses and lengths (*Clauser et al., 1969*), larger muscular activation efforts can be assumed for males than for females performing the holding task. Thus, hereto referenced muscle activity captured during landing would be consequently lower in males, which could explain the larger gender differences presented in *Kulas et al.*'s study (*2006*). Though, our findings suggest that gender itself has in comparison with previous landing familiarity only a subordinate effect on landing proficiencies and peak lumbo-pelvic flexion accelerations. *Bruton et al.* (*2013*) hereto concordantly suggested that in an average population commonly presented lower landing skill in females is drawn to their equally common lower experiences with motor tasks in the social environment, particularly in their younger age, where acquirement of motor skills is most formative (*Payne & Isaacs, 1991*). The in our study specifically undertaken segregation and comparison of gender- and familiarity-ascribed alterations on trunk muscle activity and peak flexion accelerations mostly negate an actual effect of gender on spinal stabilisation strategies and achievements in drop landings. Though, our analysis revealed that particular associations between muscular activations to the occurrence of peak lumbo-pelvic flexion accelerations seem to be gender specific. Herein Externus Obliquus activity and its co-contraction ratio with Erector Spinae were remarkably strong positively associated with peak lumbo-pelvic flexion acceleration in males but not in females. Note: that this finding was not caused by an overrepresentation of males in the landing familiar group, since the recruitment into groups of landing familiarity accounted for an equal dispersion of gender. However, the overall absence of any female specific association between trunk muscular activity and $\alpha_{FLEX[LS\text{-}PV]}$ shows, that our investigation couldn't reveal any generally valid feature of females' trunk muscular spinal impact counteraction in landings. Possibly females occasionally employed additional other trunk stabilising muscle groups (e.g. pelvic floor, diaphragm; *Kulas et al., 2006*), which were not captured by our study. Besides, the absence of associations in females might be referred to the substantially larger inter-individual variability of $\alpha_{FLEX[LS\text{-}PV]}$ in this group. This finding could be argued with the standardised landing height (*45cm*), which may have imposed subjectively higher task demands to a fair amount of females in our group, by herewith exceeding these females' previously experienced peak jumping heights (*Walsh et al., 2007; Rice et al., 2016*). The hereof posed increased task demand could have somehow overstrained these females' landing expertise and skill (*James et al., 2000; Santello, 2005; Nordin et al., 2016*). *James et al.* (*2000*) and *Nordin et al.* (*2016*) hereto showed that superelevations in landing demands and impact magnitude lead to larger variability in landing kinematics and experienced impulses. Moreover supporting the herein yielded theory of preponderantly overstraining many females' landing skills by the administered standard landing height, was the presented substantially elevated co-activity of Erector Spinae and Transversus Abdominis with Internus Obliquus in the group of females, which may signify an approach of excessive spinal stiffening. Such strategies have been shown to be most common, when provided task demands exceed the performers' retrievable skill (*Davids et al., 2003; Nordin et al., 2016*).

Drop jumps proffered slightly lower differences for $\alpha_{FLEX[LS\text{-}PV]}$ between landing familiar and unfamiliar individuals, as between genders. Despite, group differences in Transversus Abdominis & Internus Obliquus *IMPACT* activity between each respectively compared groups of alleged contrasted landing skill (familiarity and gender) were substantially accentuated in the *IMPACT* phase, but negligible during *DROP*. Herein, compared to mere drop landings, familiar individuals emphasised Transversus Abdominis & Internus Obliquus activation at drop jump's *IMPACT* by *+17%MVC* concurrently with Erector Spinae *+20%MVC*. Landing unfamiliar individuals by contrast, elevated Transversus Abdominis' & Internus Obliquus' activity by merely *+8%MVC* during *IMPACT*, whilst Erector Spinae activity at *IMPACT* increased by *+18%MVC*. Other muscles did neither show notable activity increases between drop landings and drop jumps nor between familiarity groups. This presentation may indicate that the by landing familiar individuals' acquired altered trunk activity pattern, of accentuating Transversus Abdominis & Internus Obliquus activity during the *IMPACT* phase of landing, is equally utilised in the presents of altered task demands, as yielded by drop jumps. This presented augmentation of Transversus Abdominis & Internus Obliquus activity might have been primarily evoked to adequately co-activate with the increased activity of Erector Spinae (*Krajcarski et al., 1999; Iida et al., 2012*). Elevations of

Erector Spinae's activity during *IMPACT* in drop jumps are assumed to serve the additional trunk extension effort to support the vertical acceleration of the performer's body into the requested follow-up jump (*Ambegaonkar et al., 2011; Iida et al., 2012; Prieske et al., 2013*). An adequate co-activation of Transversus Abdominis & Internus Obliquus hereto appears requisite to counterbalance Erector Spinae's extension torque (*Krajcarski et al., 1999; Iida et al., 2012*). It should be noted, that the increased activation of Erector Spinae during the *IMPACT* phase of drop jumps, seen in both: landing unfamiliar and familiar individuals, is believed to present a pre-programmed motor execution (*Ambegaonkar et al., 2011; Iida et al., 2012*) based on the apprehended task description. Whereas a proportionate abdominal co-activation, is rather assumed to rely on previously conditioned reflex activity.

Unsuspectedly, females presented, similar to landing familiar individuals, elevations of Transversus Abdominis & Internus Obliquus activity along with Erector Spinae activity during *IMPACT* in drop jump compared to drop landing (*[TrA-IO] +18%MVC, [ES] +17%MVC*); whilst males in equivalent comparisons rather accentuated their Erector Spinae activity (*[TrA-IO] +8%MVC, [ES] +25%MVC*). Notably, females hereof experienced lower peak lumbo-pelvic flexion accelerations, which hereby converged to the between landing types unchanged peak accelerations of males. Hereof it can be deduced that females ultimately gain larger trunk stability, thus experiencing diminished peak lumbar bending load, by executing a follow-up jump in immediate succession to landing.

The overall absence of any remarkable difference in peak accelerations and muscular activity between groups of familiarity or gender in *DS* landings is adoptively referred to the previously expounded subject's realisation of that task (*see chapter Effects of Landing Characteristics*).

Clinical Relevance and Implications

The spine appears to be substantially affected by landing impacts, whereof trunk responses of high trunk muscular co-activations and vigorous spine bending loads emerge collaterally to previously evinced vertical impact compressions (*DeVita & Skelly, 1992; Recknagel & Witte, 1996; Zhang et al., 2000, 2008; Hume et al., 2013; Panther & Bradshaw, 2013; Simons & Bradshaw, 2016*). The observed responses provide insights in the previously unknown integral conduct of the trunk in the absorption and dissipation of landing impact energy. Hereof it can be deduced, that the spine should not be generally considered as a single rigid element in integrally conducted landing performances; instead our findings evince that a majority of individuals permit spine segmental excursions, which imply a dissipation of energy at these spinal sectors. However, our data does not clarify if these excursions occur for the most part deliberately or accidentally. The in our observation occasionally most sudden excursions are certainly rather a result of deficient spinal stabilisation and must be appraised as vigorous bending load to the spinal tissues.

Overload and Injury.

Revealed bending accelerations emerging in landings drastically exceed the magnitudes of angular accelerations previously reported in high overload risk work places (*Marras et al., 1993*), by a *20*-fold on average and a *50*-fold on the peak expressions observed. By that exposition, our findings presently indicate that drop landings can elicit alarmingly severe spinal bending loads, which most vigorously occur in sagittal flexion of the lumbar spine. The herewith presumed accommodated rapid dorsal extension assumingly exposes individuals to a high risk of spinal soft tissue injury, such as supposed in presentations of low back pain (*Marras et al., 1993*). In unconstrained landings of lower kinetic energy potential, spinal bending accelerations and herewith assumed bending load was found to be generally humble, whilst unilateral landings from substantial landing heights appear to generate more vigorous bending loads to the lumbar spine in both sagittal and lateral flexion. Hence unilateral landings might potentially most severely hazard the spinal tissues' physical integrity. Despite, highest spinal load is elicited in landings, when performers' proficiencies were overstrained by the landing demands. In those landing exertions highest peak spine bending accelerations were frequently found. Alongside, few landing performances under excessive demands led to a near absence of any bending acceleration, suggesting that less skilful individuals approached the landing impact with a completely stiffened spine. Such approaches of superelevated spinal stiffening can detrimentally create large compression loads to the spinal column (*NIOSH, 1981; Recknagel & Witte, 1996*). Hence our findings highlight that spine challenging vigour of landings must be appraised as the multi-factorial consequence of each extrinsic landing demand and the performer's individual intrinsic landing skill.

Bone Formation Effects.

By the objective of deliberately applying extensive loads to the human skeleton, high impact magnitudes of jumps and landings have found frequent utilisation in bone material modulation approaches. A large body of bone formation research has evinced that bone responds with mass and density increases when high muscular strain or skeletal stress loads are administered (*Kohrt et al., 2004, 2009*). However, these studies have yet inconclusively showcased the lumbar spine, to present either lower (*Bassey et al., 1998; MacKelvie et al., 2003; Weeks et al., 2008; Gunter et al., 2008; Bolton et al., 2012*), similar (*MacKelvie et al., 2002; Kontulainen et al., 2002, 2004; Kato et al., 2006; Meyer et al., 2011*), or more significant adaptions to vertical jump training regimen when compared to the femur (*Fuchs et al., 2001; Vicente-Rodriguez et al., 2004; Winters-Stone et al., 2011; Maggio et al., 2012; Saarto et al., 2012; Hinton et al., 2015*). Most studies herein assessed effects of long term jumping or landing training, determining their load intensity by means of ground reaction forces. Our findings propose that the adaptive effect on the lumbar vertebrae is not solely dependent on the registered ground reaction forces, but rather substantially altered by intrinsic load modifying factors in landings. Our study hereto most dominantly revealed a task dependent significant influence of trunk muscular activity deployment modifying the onset of peak spinal bending accelerations. Those should be consulted for the understanding of bone formation effects, when a load application for elicitation of exercise effects is desired. Hence, it can be practically suggested to monitor lumbar spine accelerations, or aggravate landing demands, to improve bone formation outcomes in thereupon targeting exercise interventions.

TRAINING.

As our study was limited to a cross-sectional comparison of landing familiar vs. landing unfamiliar individuals, we could not expose the actual origination of the apparently improved trunk muscular stabilisation pattern. An implementation of controlled longitudinal intervention studies, wherein divergent landing executions are administered to and practiced by matched landing recruits, is recommended to disclose the predominant causes for trunk muscular adaptions. Such intervention would be furthermore assumed to reveal much sharper distinctions between trunk muscular activation patterns and lumbar acceleration onsets. Furthermore the effect of specifically trained and practiced volitional spine stabilisation techniques (*Haddas et al., 2016*) on sudden lumbar bending load should be persistently investigated in landing performances. Such should be fortified by feed-back methods, which have previously shown to be effective in reducing peak ground reaction forces in landing (*Ericksen et al., 2013*). Suchlike intervention, by our presumption, would moreover substantiate differences in sudden lumbar bending loads and possibly help researchers, clinicians and trainers to understand and incorporate spine health preserving landing pattern in the future. Moreover, as frequent occurrences of spinal bending accelerations have been formerly shown to entail the risk of promoting spinal overload disorders (*Garg & Moore,* *1992; Marras et al., 1993; Yamamoto, 1997*), frequently performed landings within sport- and training routines may entail an equivalent potential of fatiguing and consequently jeopardising the spinal tissue integrity. Hereto, our research recommends the implementation of further studies, pursuing on the effects of repetitive landing bouts. Overweight should be greatly regarded in landing exertions, since it appears to drastically increase experienced peak lumbar bending loads. However, receivable prove for this prevailing assumption has to come by future studies.

LIMITATIONS

LANDING DEMANDS AND SUBJECT RECRUITMENT.

Recruitment of subjects to both our research trials might have derogatorily affected the overall presented large variability of observed peak spine segmental angular accelerations. Hereto our first study's results were constraint to a relatively small heterogeneous cohort with concomitant large differences in anthropometrics and presumed overall low but diverging landing skill. Unfortunately, even in our second, larger and anthropometrically more homogeneous cohort, as within groups of presumed similar landing skill, $\alpha_{FLEX[LS\text{-}PV]}$ remained largely variable between and within individuals. Despite a presupposed generally high variability of landing performances between individuals (*James et al., 2000; Nordin et al., 2016; Nordin & Dufek, 2017*), large inter-individual variability of our assessed trunk measures, particularly within the landing familiar and female group, suggests our studies' methodological flaws in the assignment of individuals into categorical groups and to deliberately prescribed landing task demands. Firstly hereto, each landing task examined in our landing protocols was implemented from metrically standardised heights. Such may have engendered substantial discrepancies in subjectively experienced landing demands between tested individuals, which most likely caused excessive demands most frequently on females landing from *45cm* of height (*Walsh et al., 2007; Rice et al., 2016*). Subjectively experienced task demands by standardised landing heights have been previously shown to largely alter landing performances and execution variabilities (*James et al., 2000; Davids et al., 2003; Nordin et al., 2016; Nordin & Dufek, 2017*). Thus testing standard landing heights does not seem to be the most appropriate choice in comparing individuals, even though it provides the benefit of entailing most comparable gravitational potential energy. Other research groups have, with particular regards to previous findings on landing's variability, compared individuals on landing heights relative to their individual peak jump height (*James et al., 2000; Nordin et al., 2016; Nordin & Dufek, 2017*). Hence, application of verified capability related landing heights might have led to much lower variability presentations, probably for both inter- (*James et al., 2000*) and intra-individual diversity (*Davids et al., 2003*) of peak spine segmental angular acceleration outcomes in our study; this should be further considered in succeeding landing research studies.

A second aspect, suspected to inadvertently increasing inter-individual variability in our recruited groups of landing familiarity was an apparently impaired sensitivity for skill accreditation in our recruitment strategy. Landing familiarity assignment in our study was determined from any retrospect landing experiences in the past ten years preceding our landing trials. This approach might have neglected some individuals' durable landing motor skills, potentially acquired at younger maturing ages (*Payne & Isaacs 1991*), and besides assigned few individuals to the landing familiar group, who never exerted any landing on similar vigour to those tested in our landing protocol. Thus, on an individual observation of performances within both groups of familiarity, few individuals in each group stood out with intra-individually consistent respective remarkably low or high peak accelerations measured. From reviewing these particular individuals' previous landing experience reports: several unfamiliar individuals, scoring at low peak accelerations, reported extensive landing experience from younger ages, whilst other categorised landing familiar individuals, who demonstrated high peak accelerations, reported their participation in physical activity involving rather moderate landing exertions. Hence, retrospection of sole landing quantities in a determined past timespan does apparently not provide sufficient sensitivity to distinctively segregate landing skilled from landing unskilled performers. The hereof potentially resulting adverse increase of inter-individual variability of our trunk measures within the assessed groups, has assumingly unfavourably diminished the discriminatory power of most aspects in our between group-analyses.

KINEMATIC DATA.

The ultimately aggregated high intra-individual variability of our trunk measures likely originated from the subordinate lower extremity impact damping performance, which expectedly strongly modifies the remains of impact at the pelvis (*Devita & Skelly, 1992; Zhang et al. 2000*). Alterations on the pelvic impact are supposed to elicit altered reflexive trunk muscular deployment, hence affecting the hereupon depending peak spine segmental bending accelerations. Hence, lower limb damping performance variability in drop landings appear of large significance for the trunk measures we assessed. A capture of lower extremity kinematic performance may have thus provided auxiliary information about its influence on peak lumbar bending load's variability and its magnitudes. Particularly in consideration of a believed integral motor conduction of impact dissipation (*Santello, 2005*), variations in spinal kinematics might be explicable by functionally interlinked alterations in lower limb kinematic behaviour. However, our study neglected the assessment or standardisation of lower limb performances for two practical reasons. Herein firstly our study avoided an unnatural standardisation of technical landing executions, upon the aim of exposing veridical in-vivo representations of individual peak spinal bending accelerations; while secondly a limitation of the volume of parametrised data in our research trials was expedient, due to the large amount of already versatilely captured influential parameter data.

By the intention of condensing our measures to the finite lumbo-pelvic flexion acceleration, position measures of the trunk and the spinal segments were also neglected. Though, trunk posture has been previously presented to influence the activity of trunk muscles and spinal load in drop landing and sudden spine perturbations (*Recknagel & Witte, 1996; Zhang ey al., 2000; Granata & Rogers, 2007*). However, while the overall trunk's forward inclination was likely disparate between tested individuals, sectional spine segmental angles, at which peak spine segmental angular accelerations were captured, were likely consistently held in ranges near an erect posture. Our study however provides, due to the lack of data assessment, no verification on this assumption. Assessment or standardisation for such kinematic parameters might expound further insights into the backgrounds of our observations.

As previously denoted, peak spine segmental angular accelerations (α_{MAX} and $\alpha_{FLEX[LS\text{-}PV]}$) demonstrated only a limited sensitivity to detect ground impact affiliated spinal bending load, which was rather restraint to more vigorous landings causing larger loads. Hence, the measure of peak spine segmental angular accelerations should exclusively be further utilised in landings, where such high loads are certainly assumed.

EMG Recordings.

Normalisation of electro-myographic measures to task unrelated, maximal voluntary contraction tests always accommodates a risk of falsely normalising to rather submaximal muscular activity (*Konrad, 2005*). Hereto, specifically the overrepresented Erector Spinae activity displayed in our study was likely resulting from such normalisation to submaximal activation yielded from its preceding *MVC*-test trial. This can be assumed, since Erector Spinae's *MVC*-test was exclusively conducted without additional resistance from the investigator acting against the individual's voluntary contraction. Though, testing Erector Spinae this way was a decision, made upon previous experience, showing that an application of the investigator's maximal resistance against the tested individuals' thighs and dorsally lifted shoulder girdle in the given prone position could not be assured, due to the long lever necessary to fix both distal anatomical structures. However, despite the thereby in our approach resulting relatively higher values of Erector Spinae's *%MVC* activity compared to other muscles, our analyses are not believed to be further confounded in their overall presentation of trunk muscle activity pattern.

Further normalisation errors in our study could have occurred by any retention of the individuals spent effort in the testing situation, or by coordinative difficulties in the maximal contraction task execution. The most confounding effect hereof accrues by inter-individual disparate capacities to correctly execute the reference task. Our conducted evaluation of different trunk muscles was carried out by manually assisted breaking tests recommended by *Konrad* (*2005*). Throughout the implementation of the normalisation tasks, the observed bio-feed-back monitoring showcased, that several individuals had difficulties predominately employing the specific muscle, to which each respective testing trial was directed. This coordinative difficulty is likely ascribed to the required interplay of multiple trunk muscles to execute any desired trunk movement, whereof individuals, which were not previously familiarised with arbitrary particular trunk muscle activations, may have potentially employed additional other motion supporting muscles within our reference testing. These inter-individual differences could have had a confounding effect on the inter-individual variability of muscular activity and consequently on the associations of muscular activity to peak lumbo-pelvic flexion accelerations. However, the degree to which our findings were hereby potentially influenced is uncertain.

Conclusion

The trunk's participation in the impact dissipation of drop landings can occasionally result in high peak spine segmental flexion accelerations, which must be considered to entail concerning high spinal bending loads, which could overstrain spinal soft tissue. Greatest flexion accelerations to the spine during bipedal drop landings occur as sudden lumbar flexion, likely constituted by the lumbar spines larger compliance and the predominant antero-inferior force vector. Peak lumbo-pelvic flexion accelerations appear to be highly depending on both: the landing's extrinsic kinetic energy potential and the performer's utilisation of impact attenuating landing skill. Herein meticulously composed trunk muscular activity appears to play a crucial role in impact load alleviating spinal stabilisation. Superior spinal stabilisation against lumbar bending load is apparently warranted by moderate preparatory spinal stiffening and an accurate synergistic elicitation of conditioned reflexes of several trunk muscles. Herein, co-activations of Erector Spinae and Transversus Abdominis & Internus Obliquus seemingly provide the greatest contribution to lumbo-pelvic and spinal stability. Superelevated kinetic energy potential and unfamiliar landing constraints might lead to disruptions of the individually composed spine protecting trunk muscular activity pattern. As a consequence, substantially increased lumbar bending or compression loads could be evoked, from approaching the landing impact with excessive spinal stiffening; both possible herein directed load shifts may be hazardous to the lumbar spinal tissue.

Previous landing familiarity constitutes the preponderant subject characteristic yielding superior trunk muscular activation pattern, thus reducing lumbar impact loads in drop landings. Males appear to present moderate and more organised trunk muscular pre-activation pattern, whilst females seem to employ greater trunk muscular activity to stabilise the spine against identical extrinsic landing conditions. Females with lower landing skill, may benefit from an execution of a contrived imminent follow up task succeeding touch down, such as drop jumping, which herein features slightly diminished peak lumbar flexion accelerations. This effect might be evoked by a systemically elevated muscular stiffness serving the follow up task implementation. Such effect is diminished when the precocious conception of the successive task is precluded.

Functional and clinical effects of specific landing training, performers' bodyweight, and exertion frequencies should be furthermore investigated in regard to spinal bending loads from vigorous landings.

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APPENDIX 1

Table A1.1: Subject characteristics

ID	Gender	Weight (kg)	Height (m)	Age (yrs)	Physical activity involvement
01	male	67	1.68	24.0	Fencing
02	male	88	1.88	27.9	Canoe flatwater-sprint
03	female	43	1.52	24.9	Cycling, Hiking
04	male	77	1.77	30.2	Soccer
05	male	82	1.80	27.5	Basketball, Soccer, Running
06	male	70	1.90	26.5	Soccer, Beach-volleyball
07	male	76	1.67	28.8	Running, Strength exercise
08	female	74	1.79	34.3	Fitness
09	female	53	1.68	26.0	Cycling, Tennis, Running
10	female	69	1.64	26.2	Zumba
11	female	51	1.54	29.4	/
12	male	90	1.85	31.4	Cycling
13	male	82	1.87	38.6	Cycling
14	female	75	1.79	31.1	Cycling
15	male	72	1.78	27.9	Cycling, Fitness
16	female	56	1.67	29.9	Cycling, Running
17	female	58	1.65	26.7	Horseback riding

grey shaded subjects were excluded from data analysis

Table A1.2: Spearman's Rank Correlation *(r_s)* between *t-IGC<t-α_{MAX}(%)* to α_{MAX}

	r_s *sig.*	r_s *sig.*	r_s *sig.*	r_s *sig.*
Type (df)	**BL** (48)	**BF** (24)	**OH** (48)	**UL** (48)
	.59 *.000*	.72 *.000*	.58 *.000*	.67 *.000*
Height (df)	**15** (48)	**30** (48)	**45** (36)	**60** (36)
	.67 *.000*	.65 *.000*	.67 *.000*	.63 *.000*
Joint (df)	**UTS-LTS** (42)	**LTS-LS** (42)	**LS-PV** (42)	**PV-GF** (42)
	.76 *.000*	.11 *.470*	.85 *.000*	.59 *.000*
Motion (df)	**FLEX** (56)	**LAT** (56)	**ROT** (56)	
	.54 *.000*	.78 *.000*	.66 *.000*	
ALL	(168)			
	.68 *.000*			

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Table A1.3: Inter-individual Variability (***CV_{AV}%***) of α_{MAX} overall and of each *Joint*Motion**landing *Type*Height* combination

			PV-GF			LS-PV			LTS-LS			UTS-LTS		
		Mean	FLEX	LAT	ROT	FLEX	LAT	ROT	FLEX	LAT	ROT	FLEX	LAT	ROT
Mean		**66**	**61**	**85**	**59**	**73**	**82**	**60**	**62**	**65**	**62**	**62**	**57**	**64**
BL	*15*	**73**	62	105	83	68	79	59	58	117	58	65	62	66
	30	**72**	68	100	53	78	84	56	74	70	82	68	76	56
	45	**68**	74	90	70	80	78	52	79	56	51	75	53	58
	60	**59**	42	68	60	49	63	77	53	61	46	81	58	45
BF	*15*	**55**	36	80	61	52	71	41	72	54	56	53	51	37
	30	**89**	59	133	147	76	134	91	72	78	90	55	64	64
OH	*15*	**67**	65	109	41	80	81	62	63	77	85	52	51	39
	30	**70**	67	86	53	81	78	50	81	85	65	78	44	68
	45	**67**	58	99	52	74	88	62	50	73	64	65	64	58
	60	**70**	57	93	43	58	80	60	46	68	62	56	64	154
UL	*15*	**63**	66	76	51	79	91	60	50	51	56	76	49	57
	30	**67**	85	66	51	97	89	61	53	51	57	67	56	68
	45	**62**	95	58	42	97	82	54	55	43	47	49	61	66
	60	**42**	22	27	18	51	46	60	56	25	55	36	44	61

Table A1.4: Intra-individual Variability by (***CV%***) of α_{MAX} across landing *Types*Heights* and *Joints*Motions*

				Subject -ID																
			Subjects Mean	*01*	*02*	*03*	*04*	*05*	*06*	*07*	*08*	*09*	*10*	*11*	*12*	*13*	*14*	*15*	*16*	*17*
Landings / Motions Mean			**36**	**32**	**43**	**32**	**27**	**46**	**28**	**36**	**29**	**26**	**32**	**41**	**27**	**38**	**42**	**49**	**45**	**31**
Landing	*BL*	*15*	**41**	25	52	34	38	62	40	33		31	45	42	27	42	55	61	30	31
		30	**37**	29	32	38	22	54	40	36	29	27	35	41	28	39	51	51	43	32
		45	**35**	24	44	30	17	63	20	37	28	27	33	47	28	39	34	47	43	38
		60	**35**	34	34		26	49	27	36					25	40		44		
	BF	*15*	**34**	34	32	36	19	29	33	34	34	28	41	44	14	32	41	36	58	31
		30	**38**	75	47	31	29	42	30	39	19	21	31	44	34	38	36	58	34	33
	OH	*15*	**39**	26	47	33	30	55	34	43	34	32	33	43	29	44	61	57	51	16
		30	**39**	37	36	30	18	46	24	29	29	44	37	37	27	44	64	57	57	43
		45	**38**	34	45	29	28	50	23	36	15	24	38	46	36	30	41	57	68	48
		60	**39**	25	52		47	48	27	34					26	44		53		
	UL	*15*	**31**	28	37	33	31	37	27	29	52	15	22	32	24	39	20	48	34	18
		30	**31**	24	52	25	31	34	20	41	25	22	21	38	31	34	21	48	38	22
		45	**32**	21	48	31	23	34	22	33	25	17	21	41	28	36	37	42	41	35
		60	**31**				20	37		38					22	35		32		
Motion	*UTS-LTS*	*FLEX*	**34**	18	39	44	33	42	33	35	21	32	22	43	23	24	35	51	50	39
		LAT	**39**	36	40	32	41	43	30	46	39	28	42	41	24	37	46	41	48	40
		ROT	**35**	30	40	34	38	42	17	42	24	19	43	47	24	37	43	35	44	40
	LTS-LS	*FLEX*	**29**	25	42	28	13	41	30	35	24	16	17	25	20	38	42	41	33	23
		LAT	**35**	26	46	29	25	45	31	37	35	25	41	44	27	39	43	46	34	31
		ROT	**39**	32	52	36	29	43	26	34	45	27	45	37	29	49	49	45	55	35
	LS-PV	*FLEX*	**32**	30	44	20	20	37	23	29	17	25	15	29	31	45	38	59	43	22
		LAT	**40**	39	47	35	27	58	31	37	28	34	33	50	36	38	44	70	46	24
		ROT	**39**	40	46	30	34	58	33	41	34	24	37	44	29	38	44	42	35	42
	PV-GF	*FLEX*	**28**	31	45	22	15	33	25	19	18	22	15	37	18	33	29	46	47	20
		LAT	**41**	40	38	33	28	65	34	39	28	33	46	49	35	38	49	68	60	31
		ROT	**37**	38	37	38	24	44	25	35	36	31	34	50	31	44	41	48	47	32

Table A1.5: Between-day Reliability Analysis of α_{MAX} (*rad/s*^{2}) in *M1, M2* by descriptive (*Mean±SD, Min., Max.*) and inferential statistics (*ICC 2.1, TRV%* and *Bland & Altman Analysis*) for distinct *Joint*Motions* (arranged by *ICC 2.1*)

	Mean ± SD (M1)	***Mean ± SD (M2)***	***Min.***	***Max.***	***TRV %***	***Bias ± LoA***	***ICC 2.1***
FLEX*[UTS-LTS]*	18 ±10	19 ±11	1	135	35	1 ±14	.811
FLEX*[LS-PV]*	87 ±57	104 ±57	9	506	46	18 ±82	.759
FLEX*[PV-GF]*	83 ±46	97 ±43	2	478	39	16 ±69	.723
LAT*[LS-PV]*	60 ±35	79 ±36	4	936	32	16 ±49	.722
LAT*[PV-GF]*	59 ±34	65 ±26	3	551	27	3 ±55	.669
ROT*[UTS-LTS]*	23 ±12	22 ±13	3	539	28	-1 ±20	.665
LAT*[LTS-LS]*	34 ±15	36 ±12	2	215	27	0 ±24	.632
ROT*[PV-GF]*	34 ±12	36 ±13	1	354	27	2 ±24	.609
ROT*[LTS-LS]*	41 ±21	41 ±19	2	380	31	-4 ±34	.578
FLEX*[LTS-LS]*	63 ±34	64 ±22	2	304	39	-6 ±56	.569
ROT*[LS-PV]*	38 ±20	37 ±12	4	343	29	-3 ±38	.396
LAT*[UTS-LTS]*	26 ±13	24 ±7	2	256	33	-2 ±22	.317

Table A1.6: Between-day Reliability Analysis of α_{MAX} (*rad/s*^{2}) in *M1, M2* by descriptive (*Mean±SD, Min., Max.*) and inferential statistics (*ICC 2.1, TRV%* and *Bland & Altman Analysis*) for distinct landing *Type*Heights* (arranged by *ICC 2.1*)

	Mean ± SD (M1)	***Mean ± SD (M2)***	***Min.***	***Max.***	***TRV %***	***Bias ± LoA***	***ICC 2.1***
UL60	114 ±35	131 ±64	16	361	30	17 ±76	.738
UL15	41 ±20	49 ±19	2	292	24	7 ±29	.736
UL30	61 ±30	76 ±26	3	416	31	15 ±44	.675
BL30	35 ±23	32 ±11	2	269	48	-3 ±33	.659
UL45	83 ±40	111 ±41	7	509	40	28 ±59	.648
OH15	31 ±18	34 ±13	3	266	39	3 ±30	.643
BL45	38 ±22	38 ±12	1	353	37	0 ±36	.613
OH30	40 ±24	45 ±19	4	312	43	5 ±43	.582
OH45	48 ±27	53 ±26	3	391	42	5 ±56	.515
OH60	48 ±24	48 ±15	3	539	45	1 ±44	.434
BF15	25 ±9	26 ±8	3	171	33	1 ±21	.321
BL15	25 ±15	26 ±8	1	192	48	1 ±32	.285
BF30	39 ±28	39 ±17	2	600	56	0 ±78	.217
BL60	39 ±16	36 ±9	4	303	49	-3 ±41	.004

Table A1.7.: Subjects' individual α_{MAX} (*rad/s*^{2}; *Mean±SD*) on each *Joint*Motion* on average over all landing *Types* and *Heights*

	PV-GF			***LS-PV***			*LTS-LS*			*UTS-LTS*		
Subjects:	*FLEX*	*LAT*	*ROT*	***FLEX***	*LAT*	*ROT*	*FLEX*	*LAT*	*ROT*	*FLEX*	*LAT*	*ROT*
ALL	83 ± 46	59 ± 34	34 ± 12	**87 ± 57**	60 ± 35	38 ± 20	63 ± 34	34 ± 15	41 ± 21	18 ± 10	26 ± 13	23 ± 12
1	71 ± 77	53 ± 93	46 ± 57	**76 ± 86**	68 ± 106	46 ± 30	65 ± 30	28 ± 18	35 ± 24	45 ± 15	30 ± 16	19 ± 9
2	31 ± 16	27 ± 22	13 ± 8	**33 ± 12**	27 ± 22	19 ± 12	32 ± 18	22 ± 14	21 ± 17	9 ± 4	15 ± 9	17 ± 12
3	63 ± 29	40 ± 20	21 ± 15	**69 ± 31**	39 ± 19	26 ± 13	90 ± 37	33 ± 25	30 ± 21	13 ± 6	23 ± 17	14 ± 6
4	108 ± 41	52 ± 44	40 ± 29	**101 ± 52**	34 ± 20	41 ± 25	67 ± 31	49 ± 47	48 ± 36	20 ± 14	47 ± 45	50 ± 75
5	128 ± 63	95 ± 70	41 ± 23	**124 ± 70**	105 ± 64	97 ± 72	109 ± 60	67 ± 32	88 ± 83	28 ± 18	57 ± 67	36 ± 19
6	46 ± 21	23 ± 23	16 ± 12	**42 ± 26**	24 ± 20	13 ± 6	25 ± 14	12 ± 7	18 ± 16	12 ± 6	8 ± 6	11 ± 8
7	104 ± 46	34 ± 27	48 ± 45	**86 ± 48**	36 ± 24	33 ± 27	38 ± 30	25 ± 34	29 ± 32	15 ± 10	34 ± 44	20 ± 18
9	64 ± 34	45 ± 60	27 ± 25	**61 ± 54**	45 ± 71	33 ± 17	39 ± 16	26 ± 26	30 ± 22	11 ± 5	14 ± 8	20 ± 12
10	221 ± 97	155 ± 91	56 ± 38	**266 ± 93**	152 ± 128	45 ± 33	79 ± 37	47 ± 30	40 ± 32	25 ± 13	24 ± 14	37 ± 25
11	73 ± 41	73 ± 45	27 ± 18	**114 ± 41**	84 ± 48	45 ± 17	149 ± 72	47 ± 28	74 ± 43	24 ± 9	31 ± 16	37 ± 24
12	85 ± 33	61 ± 58	33 ± 36	**86 ± 51**	66 ± 60	31 ± 28	77 ± 40	33 ± 43	59 ± 78	22 ± 15	29 ± 43	36 ± 44
13	70 ± 26	35 ± 34	33 ± 38	**61 ± 22**	38 ± 31	29 ± 32	47 ± 40	35 ± 46	55 ± 73	19 ± 11	28 ± 26	15 ± 12
14	93 ± 41	66 ± 66	25 ± 16	**99 ± 42**	79 ± 74	28 ± 15	38 ± 17	25 ± 21	27 ± 20	11 ± 5	17 ± 12	13 ± 7
15	41 ± 22	32 ± 45	34 ± 37	**30 ± 17**	35 ± 44	26 ± 21	27 ± 12	17 ± 21	18 ± 17	9 ± 9	18 ± 23	11 ± 6
16	49 ± 32	89 ± 61	45 ± 37	**50 ± 31**	69 ± 46	51 ± 41	61 ± 30	50 ± 50	47 ± 51	9 ± 5	20 ± 20	17 ± 9

boxed subject (**10**) presented conspicuous overweight

Table A1.8: Cohort (*Mean±SD*) of α_{MAX} (*rad/s*^{2}) from each landing *Type*Height* in each *Joint*Motion*; with highlighted contrasts to $\alpha_{FLEX[LS\text{-}PV]}$ (grey shaded)

		PV-GF			***LS-PV***			*LTS-LS*			*UTS-LTS*		
		FLEX	*LAT*	*ROT*	***FLEX***	*LAT*	*ROT*	*FLEX*	*LAT*	*ROT*	*FLEX*	*LAT*	*ROT*
ALL		83 ±46	59 ±34	34 ±12	**87 ±57**	60 ±35*	38 ±20*	63 ±34	34 ±15	41 ±21	18 ±10$^{\#}$	26 ±13	23 ±12
BL	***15***	50 ±31	29 ±30	16 ±13	**52 ±35**	31 ±24*	19 ±11*	31 ±18	19 ±22	17 ±10	11 ±7$^{\#}$	14 ±9	15 ±10
	30	101 ±68	36 ±36	18 ±10	**88 ±68**	40 ±34*	27 ±15*	45 ±33$^{\#}$	18 ±12	27 ±22	13 ±9$^{\#}$	15 ±11	17 ±9
	45	85 ±63	43 ±39	20 ±14	**87 ±69**	43 ±33*	27 ±14*	58 ±46$^{\#}$	19 ±10	20 ±10	15 ±11$^{\#}$	15 ±8	20 ±12
	60	98 ±41	30 ±20	21 ±13	**94 ±46**	41 ±26*	30 ±23*	56 ±30$^{\#}$	17 ±11	25 ±12	20 ±16$^{\#}$	17 ±10	18 ±8
BF	***15***	62 ±22	26 ±21	17 ±10	**59 ±31**	29 ±20*	17 ±7*	30 ±22$^{\#}$	14 ±8	15 ±9	11 ±6$^{\#}$	10 ±5	13 ±5
	30	95 ±56	50 ±67	30 ±44	**91 ±69**	51 ±69*	30 ±27*	46 ±33$^{\#}$	18 ±14	24 ±22	14 ±8$^{\#}$	15 ±10	17 ±11
OH	***15***	68 ±44	32 ±35	17 ±7	**77 ±62**	29 ±23*	23 ±14*	48 ±30	17 ±13	18 ±15	15 ±8$^{\#}$	13 ±7	13 ±5
	30	86 ±58	34 ±29	23 ±12	**95 ±77**	38 ±30*	27 ±14*	74 ±60	26 ±22	22 ±14	20 ±16$^{\#}$	15 ±6	17 ±11
	45	108 ±63	47 ±47	27 ±14	**110 ±81**	48 ±42*	33 ±20*	91 ±46	27 ±20	26 ±17	22 ±14$^{\#}$	20 ±13	18 ±11
	60	111 ±63	32 ±29	28 ±12	**107 ±62**	36 ±29*	35 ±21*	95 ±43	24 ±17	23 ±14	25 ±14$^{\#}$	24 ±15	34 ±52
UL	***15***	51 ±33	67 ±51	33 ±17	**56 ±44**	62 ±57	36 ±21	45 ±22	41 ±21	53 ±29	15 ±11$^{\#}$	21 ±11	19 ±11
	30	73 ±63	91 ±60	53 ±27	**76 ±74**	89 ±79	54 ±33	63 ±34	58 ±29	85 ±48	17 ±12$^{\#}$	38 ±21	30 ±21
	45	98 ±94	126 ±73	74 ±31	**105 ±102**	126 ±103	71 ±38	84 ±47	83 ±36	99 ±47	22 ±11$^{\#}$	62 ±38	49 ±32
	60	100 ±22	151 ±41	116 ±21	**100 ±51**	140 ±64*	113 ±69	113 ±63	118 ±30	163 ±90	47 ±17	135 ±59	75 ±46

* = sig. diff. from *FLEX*; $^{\#}$ = sig. diff from *LS-PV* ($p < 0.05$)

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Table A1.9: Cohort (*Mean±SD*) of *GRF_N* overall and in each landing *Type*Height*

	15	30	45	60	ALL
BL	1340 ±362	1765 ±471	2042 ±485	2507 ±645	1850 ±485
BF	1376 ±295	1868 ±384			1622 ±316
OH	1432 ±334	1880 ±486	2213 ±503	2785 ±703	1963 ±505
UL	2105 ±418	2780 ±666	3527 ±774	5201 ±1150	2967 ±827
ALL	**1569 ±314**	**2082 ±447**	**2557 ±581**	**3159 ±1003**	**2150 ±545**

Appendix 2

Table A2.1: Subject characteristics and individual landing experience

ID	Gender	Weight (kg)	Height (m)	Age (yrs)	Landing experience *Discipline / Duration / Dating*	*Familiarity group (1, 2)*
01	female	63	1.62	27	Hip Hop dance / 1 year / > 0.7 years Fitness exercise / 0.5 years / currently	*1*
02	male	86	1.83	29	Soccer / 10 years / > 6 years Basketball / 12 years / currently Fitness exercise / 45 years / currently	*1*
03	female	54	1.62	30	Modern Jazz Dance / 11 years / > 9 years	*1*
04	female	57	1.69	27	Fitness exercise / 0.5 years / > currently Soccer / 13 years / > 2 years	*1*
05	male	75	1.80	25	Volleyball / 10 years / currently Track & Field / 10 years / > 10 years	*1*
06	male	73	1.74	31	Volleyball / 1 year / > 8 years	*2*
07	female	58	1.61	29	Handball / 7 years / > 11 years Fitness exercise / 0.2 years / currently	*1*
08	male	90	1.82	32	Basketball / 16 years / currently Track & Field / 3 years / > 13 years	*1*
09	female	53	1.68	22	Jumping fitness / 18 years / currently Crossfit / 3 years / > 1 year Track & Field / 2 years / > 4 years	*1*
10	female	63	1.67	27	Dance / 20 years / > 4 years Fitness exercise / 5 years / currently	*1*
11	female	52	1.60	34	Modern Jazz Dance / 23 years / currently	*1*
12	male	78	1.77	35	Volleyball / 20 years / currently Badminton / 5 years / currently Basketball / 10 years / > 10 years	*1*
13	female	60	1.61	29	Handball / 3 years / currently Track & Field / 9 years / > 13 years	*1*
14	male	77	1.76	31	-	*2*
15	male	78	1.81	33	Volleyball / 1 year / > 6 years Gymnastics / 1 years / > 6 years Fitness exercise / 2 years / currently	*1*
16	male	76	1.82	28	Basketball / 1.5 years / > 15 years Basketball / 2 years / > 4 years Fitness exercise / 0.2 years / > 0.5 years	*1*
17	male	74	1.79	31	Frisbee / 2 years / currently	*2*
18	female	58	1.61	29	Volleyball / 6 years / > 11 years	*2*
19	female	62	1.65	29	Fitness exercise / 3 years / currently	*1*
20	female	45	1.56	27	Rope Skipping / 1 year / > 11 years	*2*
21	female	57	1.67	31	Handball / 13 years / > 7 years	*1*
22	female	57	1.63	29	Volleyball / 6 years / > 1.5 years Fitness exercise / 1 year / currently	*1*
23	male	74	1.79	29	Freelethics / 2 years / > 0.2 years Basketball / 5 years / > 3 years Soccer / 20 years / > 2 years	*1*
24	female	54	1.68	30	Track & Field / 5 years / > 20 years	*2*
25	male	73	1.80	34	Badminton / 2 years / > 2 years Basketball / 9 years / > 1 year	*1*
26	female	54	1.63	36	Gymnastics / 3 years / > 23 years	*2*
27	female	58	1.62	20	Handball / 2 years / > 12 years	*2*
28	female	56	1.62	30	-	*2*
29	male	93	1.84	35	Basketball / 5 years / > 6 years Volleyball / 2 years / > 6 years	*1*
30	female	59	1.74	31	-	*2*
31	female	56	1.67	32	Track & Field / 1 year / > 7 years	*1*

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					Handball / 0.5 years / > 7 years Gymnastics / 1 year / currently Trampoline / 0.5 years / currently	
32	male	71	1.68	27	Soccer / 1 year / > 4 years Floorball / 1 year / > 3 years Fitness exercise / 0.2 years / currently	*1*
33	male	67	1.87	30	Basketball / 0.5 years / > 16 years Handball / 0.5 years / > 18 years	*2*
34	male	80	1.82	21	-	*2*
35	male	86	1.92	38	Basketball / 7 years / > 18 years	*2*
36	male	72	1.81	18	-	*2*
37	female	59	1.66	29	Ballet / 4 years / >18 years Gymnastics / 4 years / > 18 years	*2*
38	male	103	1.91	32	Soccer / 18 years / currently Kickboxing / 1.5 years / >13 years	*1*
39	female	54	1.65	28	Bouldering / 3 years / currently Skiing / 5 years / currently	*2*
40	female	80	1.71	27	-	*2*
41	female	74	1.74	26	Volleyball / 4 years / >6 years Track & Field / 4 years / >11 years	*1*
42	female	68	1.74	36	Track & Field / 11 years / >21 years	*2*
43	male	80	1.79	39	-	*2*
44	male	79	1.75	29	Basketball / 5 years / >6 years Volleyball / 16 years / currently Snowboarding / 6 years / currently	*1*
45	male	69	1.81	25	Soccer / 8 years / >11 years	*2*

grey shaded subjects were excluded from data analysis

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Table A2.2a: Inter-individual variability (***CV_{AV}%***) of $\alpha_{FLEX[LS\text{-}PV]}$, for total cohort and groups of *Gender* (*F*,*M*) and *Familiarity* (*1*,*2*)

		Total Cohort	*Female (F)*	*Male (M)*	*Familiar (1)*	*Unfamiliar (2)*
$\alpha_{FLEX[LS\text{-}PV]}$	***DL***	**56**	*63*	*46*	*69*	*38*
	DJ	**52**	*56*	*48*	*62*	*37*
	DS	**47**	*50*	*44*	*55*	*35*
	ALL	**52**	*56*	*45*	*61*	*37*

Table A2.2b: Inter-individual variability (***CV_{AV}%***) of muscular *PRE*-activity (*%MVC*) for total cohort and groups of *Gender* (*F*,*M*) and *Familiarity* (*1*,*2*)

		Total Cohort	*Female (F)*	*Male (M)*	*Familiar (1)*	*Unfamiliar (2)*
DL	***[RA]***	**36**	36	37	35	39
	[TrA-IO]	**44**	43	45	44	44
	[EOl]	**53**	36	68	55	47
	[EOr]	**51**	47	54	38	48
	[ESl]	**29**	26	31	25	30
	[ESr]	**29**	26	29	29	27
	CCI[EO|ES]	**58**	50	69	52	55
DJ	***[RA]***	**33**	34	32	37	30
	[TrA-IO]	**50**	50	48	52	45
	[EOl]	**54**	40	68	55	50
	[EOr]	**49**	47	47	37	48
	[ES-l]	**30**	26	34	31	28
	[ESr]	**27**	19	35	29	26
	CCI[EO|ES]	**57**	53	62	48	54
DS	***[RA]***	**33**	32	34	32	34
	[TrA-IO]	**49**	51	46	53	42
	[EOl]	**54**	47	62	56	47
	[EOr]	**54**	51	55	39	51
	[ESl]	**29**	23	35	24	31
	[ESr]	**32**	29	29	35	28
	CCI[EO|ES]	**60**	57	64	51	56

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Table A2.3a: Intra-individual variability (***CV%***) of $\alpha_{FLEX[LS\text{-}PV]}$ for individuals and means for total cohort (n=43), and groups of *Familiarity* (*1,2*) and *Gender* (*F,M*)

	$\alpha_{FLEX[LS\text{-}PV]}$			
Subject ID / MEANs	***DL***	***DJ***	***DS***	***ALL***
Total Cohort	**26**	**34**	**30**	**30**
Familiar (1)	29	40	34	34
Unfamiliar (2)	22	26	24	24
Females (F)	27	33	32	31
Males (M)	25	36	27	29
*Female*Familiar (F1)*				
01	22	29	35	29
03	26	14	31	24
04	15	36	60	37
07	22	23	37	27
09	28	27	45	33
10	36	37	45	39
11	19	43	23	28
13	33	32	47	38
19	41	25	17	28
21	33	87	37	52
22	41	24	36	34
31	21	25	26	24
41	42	61	29	44
*Female*Unfamiliar (F2)*				
18	29	38	41	36
20	20	33	32	28
24	24	34	29	29
26	15	20	25	20
27	32	36	22	30
28	19	30	17	22
30	28	28	25	27
37	23	27	42	31
39	33	18	19	23
42	23	37	25	28
*Male*Familiar (M1)*				
02	47	39	47	44
05	45	70	65	60
08	25	96	42	54
12	29	53	37	40
15	18	16	26	20
16	9	21	17	15
23	20	22	30	24
25	47	91	28	55
29	24	33	20	25
32	42	42	40	41
38	12	17	15	15
44	28	44	20	31
*Male*Unfamiliar (M2)*				
14	23	25	34	28
17	14	14	16	15
33	19	17	13	17
34	27	35	25	29
35	12	15	19	15
36	11	18	14	15
43	13	16	15	15
45	30	28	21	26

Table A2.3b: Intra-individual variability (***CV%***) of single and indexed muscular *PRE*-activity (*%MVC*) for individuals and means for total cohort (n=43), and groups of *Familiarity* (*1,2*) and *Gender* (*F,M*)

	DL							*DJ*							*DS*						
	single muscular PRE-activity %						*CCI %*	*single muscular PRE-activity %*						*CCI %*	*single muscular PRE-activity %*						*CCI %*
ID	*[RA]*	*[TrA-IO]*	*[EOl]*	*[EOr]*	*[ESl]*	*[ESr]*	*[EO/ES]*	*[RA]*	*[TrA-IO]*	*[EOl]*	*[EOr]*	*[ES-l]*	*[ESr]*	*[EO/ES]*	*[RA]*	*[TrA-IO]*	*[EOl]*	*[EOr]*	*[ESl]*	*[ESr]*	*[EO/ES]*
Total Cohort	**25**	**17**	**21**	**20**	**24**	**23**	**21**	**27**	**15**	**20**	**20**	**22**	**23**	**20**	**25**	**16**	**21**	**23**	**26**	**25**	**23**
Familiar (1)	24	18	21	20	25	25	22	29	14	19	20	22	24	19	24	16	22	22	27	26	24
Unfamiliar (2)	27	15	20	21	22	21	21	24	16	21	19	21	22	21	26	16	20	23	24	24	21
Females (F)	26	15	20	19	22	21	20	23	14	21	18	22	23	18	24	16	22	22	24	23	24
Males (M)	25	18	21	22	26	26	23	31	16	19	22	22	23	22	26	16	20	23	28	28	22
*Female*Familiar (F1)*																					
01	20	45	28	14	9	23	22	46	20	17	13	24	23	20	14	26	20	19	27	18	22
03	39	10	26	20	51	36	35	48	11	26	16	39	36	33	29	9	41	17	32	29	39
04	35	17	29	16	32	30	28	27	18	23	17	26	25	16	26	25	23	25	26	33	17
07	30	19	14	16	22	14	15	43	24	23	24	12	12	22	41	20	18	14	19	21	16
09	22	16	25	28	19	15	29	24	13	23	27	15	16	16	34	17	32	63	29	27	60
10	25	13	19	19	34	18	18	22	18	13	16	32	32	14	46	14	21	20	31	22	35
11	19	15	15	16	14	23	14	14	6	23	22	12	20	13	17	11	30	15	17	11	16
13	64	10	37	19	26	10	36	19	12	24	15	27	16	17	8	23	15	15	33	22	19
19	20	14	16	22	23	20	9	13	14	18	15	23	30	12	18	11	20	24	26	22	26
21	21	12	22	16	25	31	24	21	8	22	16	20	25	17	22	11	17	16	21	24	17
22	16	17	19	19	20	21	9	38	6	25	17	12	21	19	15	20	15	16	17	20	10
31	15	18	18	18	10	11	11	24	12	17	21	19	13	16	27	30	38	31	21	28	38
41	11	11	11	20	14	14	10	13	9	15	20	22	33	12	8	13	21	12	17	33	21
*Female*Unfamiliar (F2)*																					
18	34	16	30	33	8	24	22	14	22	21	35	17	24	22	41	20	22	37	23	32	21
20	37	22	20	23	25	21	26	15	12	22	14	26	31	16	20	10	18	21	19	15	22
24	53	14	13	28	19	12	24	44	11	22	25	14	23	26	52	13	19	16	12	16	20
26	9	8	21	17	29	13	17	10	10	19	5	28	26	9	8	14	21	12	24	26	19
27	19	20	19	15	22	24	14	15	21	28	20	23	15	22	14	19	12	21	22	22	18
28	12	7	16	16	23	15	22	17	16	23	21	26	27	28	14	9	23	30	29	18	36
30	27	12	17	18	28	28	20	15	12	19	11	19	25	10	26	11	20	16	20	18	18
37	23	11	20	19	19	25	24	9	16	20	12	16	13	6	20	17	15	33	28	26	21
39	17	13	13	8	21	39	19	18	9	17	13	22	19	18	22	17	22	17	19	16	18
42	20	15	18	18	14	21	13	20	11	20	15	19	20	21	31	11	19	19	40	25	22
*Male*Familiar (M1)*																					
02	21	31	23	24	30	32	25	70	16	35	19	25	27	21	28	16	19	18	45	37	21
05	30	22	34	32	25	15	35	23	15	14	15	21	14	14	39	13	30	15	31	21	20
08	36	16	20	23	44	30	33	22	17	10	27	19	12	19	12	11	13	10	35	18	17
12	20	29	26	27	27	25	25	48	8	25	28	31	24	38	51	14	24	40	29	22	48
15	32	27	13	25	29	33	23	38	16	13	19	26	41	21	24	12	18	30	25	35	22
16	13	15	28	20	34	37	26	12	11	9	21	20	21	16	24	26	29	24	39	39	23
23	27	7	15	17	14	30	15	36	11	19	18	31	34	15	22	14	15	21	22	29	14
25	22	20	18	25	39	41	29	35	28	21	32	24	27	30	24	11	22	21	42	44	33
29	25	10	15	14	17	11	11	22	14	21	25	12	19	25	25	16	19	20	19	18	20
32	15	9	14	15	25	46	10	16	11	12	19	18	33	15	16	12	16	19	21	23	13
38	7	15	18	13	24	31	27	19	19	12	13	22	22	15	18	15	11	17	41	30	21
44	27	26	21	16	22	18	29	26	15	13	24	15	14	16	14	14	16	25	20	21	24
*Male*Unfamiliar (M2)*																					
14	16	23	27	25	14	17	29	13	20	11	44	21	24	37	15	31	28	34	29	35	26
17	42	20	39	31	13	32	45	49	18	44	23	26	25	32	23	12	23	23	22	24	21
33	46	23	20	35	53	36	13	28	12	22	12	23	27	18	65	11	25	30	46	49	18
34	23	12	10	20	21	22	17	37	15	19	16	15	28	17	24	17	19	24	28	35	26
35	16	18	23	18	26	15	14	31	27	20	17	18	14	20	17	26	27	21	20	21	25
36	17	16	26	24	18	12	27	26	16	20	28	29	19	25	26	20	17	16	14	15	20
43	9	10	12	9	17	17	15	14	14	25	21	18	18	29	12	7	12	21	11	12	16
45	62	16	20	21	23	13	10	49	24	17	18	19	28	23	37	15	17	29	22	30	21

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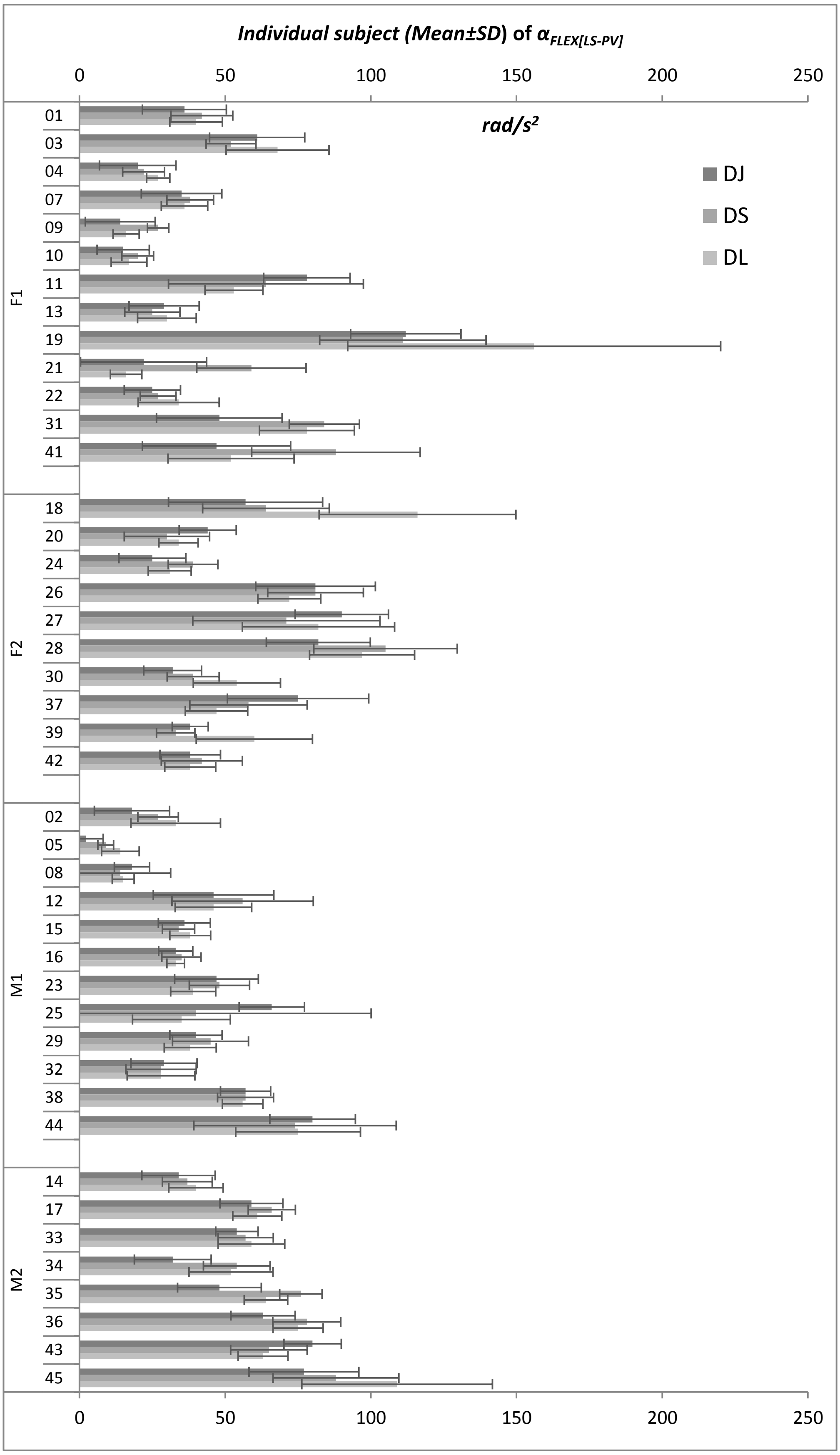

Figure A2.1: Descriptive subject's individual (*Mean ±SD*) of $\alpha_{FLEX[LS\text{-}PV]}$ (*rad/s*^{2}) in each landing *Type* (*DL, DJ, DS*) arranged into combined groups of *Gender*Familiarity* (*F1, F2, M1, M2*)

Table A2.4: Single muscular *PRE*-activity (*%MVC*) from *[RA], [TrA-IO], [EO], [ES];* (*Mean±SD*) for cohort and groups (*Familiarity, Gender*) within each, and on average of all landing *Types* (*DL, DJ, DS*)

	$[RA]_{PRE}$				$[TrA\text{-}IO]_{PRE}$				$[EO]_{PRE}$				$[ES]_{PRE}$			
	DL	***DJ***	***DS***	***ALL***	***DL***	***DJ***	***DS***	***ALL***	***DL***	***DJ***	***DS***	***ALL***	***DL***	***DJ***	***DS***	***ALL***
Familiarity																
Familiar (1)	**9.5**	**11.2** $^{*DL}_{*DS}$	**9.3**	**10.0** *	**66.4**	**74.2** $^{*DL}_{*DS}$	**65.9**	**68.9** *	**12.7**	**14.2** ^{*DS}	**12.1** $^{\#}$	**13.0** *	**39.6**	**51.4** $^{*DL}_{*DS}$	**42.2**	**44.4** *
(n=25)	±3.3	±4.1	±3	±3.5	±29.4	±38.7	±34.9	±34.3	±7	±7.7	±6.7	±7.1	±10	±15.8	±10.2	±13.2
Unfamiliar (2)	**10.4**	**11.6**	**10.7**	**10.9**	**62.2**	**65.8** ^{*DS}	**58.9**	**62.3** *	**17.2**	**18.9**	**16.7**	**17.6**	**46.0**	**57.0** $^{*DL}_{*DS}$	**49.9**	**51.0** *
(n=18)	±4	±3.5	±3.6	±3.7	±27.5	±29.7	±24.8	±27	±8	±9.5	±7.9	±8.4	±13.8	±16	±15.4	±15.5
Gender																
Female (F)	**10.4**	**12.0** $^{*DL}_{*DS}$	**10.5**	**10.9** *	**67.1**	**76.3** $^{*DL}_{*DS}$	**67.8**	**70.4** *	**14.6**	**16.2** ^{*DS}	**14.3**	**15.0** *	**44.9**	**54.3** $^{*DL}_{*DS}$	**47.0**	**48.7** *
(n=23)	±3.7	±4.1	±3.3	±3.7	±29.1	±38.3	±34.3	±33.9	±5.3	±6.5	±6.7	±6.1	±11.7	±14.2	±10.6	±12.7
Male (M)	**9.3**	**10.7**	**9.2**	**9.7** *	**61.8**	**64.3** ^{*DS}	**57.5**	**61.2** *	**14.6**	**16.1** ^{*DS}	**13.7**	**14.8** *	**39.2** ^{*DS}	**53.1** $^{*DL}_{*DS}$	**43.6**	**45.3** *
(n=20)	±3.5	±3.4	±3.2	±3.4	±27.9	±30.7	±26.4	±28.1	±9.9	±11	±8.5	±9.7	±12	±18.1	±15.5	±16.2
Cohort																
(n=43)	**9.9**	**11.4** $^{*DL}_{*DS}$	**9.9**	**10.4** *	**64.6**	**70.7** ^{*DS}	**63.0**	**66.1** *	**14.6**	**16.2** $^{*DL}_{*DS}$	**14.0**	**14.9** *	**42.3** ^{*DS}	**53.7** $^{*DL}_{*DS}$	**45.4**	**47.1** *
	±3.6	±3.8	±3.3	±3.6	±28.4	±35.1	±31	±31.5	±7.7	±8.8	±7.5	±8	±12	±16	±13.1	±14.5

***** sig. different from respective compared landing *Type* (*p<0.05*) = within subjects effect

\# sig. different from respective compared group (*p<0.05*) = between subjects effect

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Table A2.5: *CCI[EO|ES]$_{PRE}$ (%MVC);* (*Mean±SD*) for cohort and groups (*Familiarity, Gender*) within each, and on average of all landing *Types* (*DL, DJ, DS*)

	CCI[EO|ES]$_{PRE}$			
Familiarity	***DL***	***DJ***	***DS***	***ALL***
Familiar (1) *(n=25)*	**16.5** # ±8.6	**18.0** #|*DS ±8.6	**15.2** # ±7.8	**17.0** *|# ±8.3
Unfamiliar (2) *(n=18)*	**25.0** ±13.6	**27.4** *DS ±14.9	**24.3** ±13.6	**26.0** * ±13.8
Gender				
Female (F) *(n=23)*	**21.2** ±10.6	**23.4** *DS ±12.4	**20.3** ±11.6	**21.6** * ±11.5
Male (M) *(n=20)*	**18.7** ±12.9	**20.3** *DS ±12.5	**17.5** ±11.3	**18.8** * ±12.1
Cohort				
(n=43)	**20.0** ±11.6	**21.9** *DS ±12.4	**19.0** ±11.4	**20.3** * ±11.8

***** sig. different from respective compared landing *Type* (*p<0.05*) = within subjects effect

sig. different from respective compared group (*p<0.05*) = between subjects effect

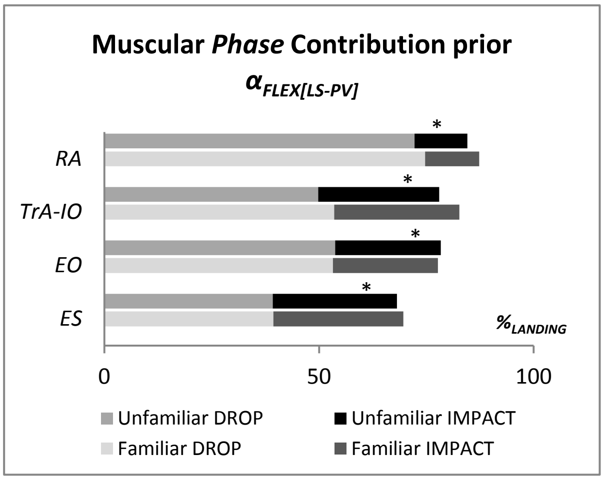

Figure A2.2: Muscular contributions (%$_{LANDING}$) contrasted by *Phase* and *Familiarity* on average of all landing *Types*;

*****sig. phase difference (*p<0.05*)

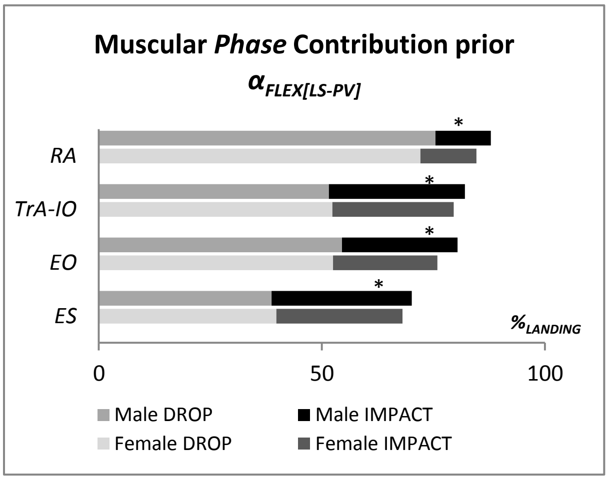

Figure A2.3: Muscular contributions (%$_{LANDING}$) contrasted by *Phase* and *Gender* on average of all landing *Types*;

*****sig. phase difference (*p<0.05*)

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Table A2.6: Descriptive (*Mean ± SD, Diff., Diff%*) and inferential comparison (*independent samples T-test*) of muscular *Phase* (*DROP, IMPACT*) contributions ($\%_{LANDING}$) between groups of *Familiarity* (*Familiar, Unfamiliar*)

	Familiar *(N=25)*		***Unfamiliar*** *(N=18)*				
	Mean	*± SD*	*Mean*	*± SD*	*abs. Diff.*	*% Diff.*	*sig. p=*
DL							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**55**	±7.3	**54**	±8.5	0.8	-1.4	.754
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**24**	±7.4	**24**	±5.9	-0.4	1.6	.858
$[ES]_{DROP}$ ($\%_{LANDING}$)	**41**	±10.6	**42**	±13.5	-1.5	3.5	.690
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**28**	±7.2	**28**	±6.7	0.4	-1.5	.845
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**55**	±7	**51**	±10.4	4.1	-7.3	.134
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**28**	±5.9	**28**	±5.9	-0.1	0.2	.975
$[RA]_{DROP}$ ($\%_{LANDING}$)	**75**	±11.3	**72**	±12.7	3.0	-4.0	.423
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**12**	±5.5	**12**	±4.5	0.0	0.1	.995
DJ							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**51**	±8.1	**53**	±6.5	-1.3	2.4	.582
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**28**	±9.6	**25**	±6.2	2.4	-8.8	.320
$[ES]_{DROP}$ ($\%_{LANDING}$)	**41**	±8.3	**39**	±10.1	1.6	-4.0	.566
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**32**	±7.2	**29**	±5.9	3.3	-10.4	.115
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**51**	±7.5	**48**	±9.4	3.1	-6.0	.243
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**32**	±7.3	**29**	±5.8	3.2	-10.1	.128
$[RA]_{DROP}$ ($\%_{LANDING}$)	**75**	±10.6	**73**	±13.4	2.8	-3.7	.448
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**14**	±6.1	**13**	±6.2	0.3	-2.5	.858
DS							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**53**	±7.8	**52**	±8.3	0.8	-1.5	.755
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**23**	±7.3	**24**	±6.8	-1.2	4.8	.599
$[ES]_{DROP}$ ($\%_{LANDING}$)	**38**	±8.3	**37**	±10.3	0.6	-1.6	.831
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**30**	±6.3	**30**	±7.6	0.3	-1.0	.888
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**54**	±6.5	**50**	±9.6	3.9	-7.3	.116
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**28**	±6.2	**28**	±5.8	-0.1	0.3	.969
$[RA]_{DROP}$ ($\%_{LANDING}$)	**74**	±10.6	**72**	±14	1.8	-2.4	.634
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**12**	±5.7	**12**	±6.1	0.5	-3.7	.806

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Table A2.7: Descriptive (*Mean ± SD, Diff., Diff%*) and inferential comparison (*independent samples T-test*) of muscular *Phase* (*DROP, IMPACT*) contributions *(*$\%_{LANDING}$) between groups of *Gender* (*Female, Male*)

	Females *(N=23)*		**Males** *(N=20)*				
	Mean	*± SD*	*Mean*	*± SD*	*abs. Diff.*	*% Diff.*	*sig. p=*
DL							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**53**	±7.5	**57**	±7	4.0	7.0	.108
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**22**	±7	**24**	±5.6	2.0	8.3	.297
$[ES]_{DROP}$ ($\%_{LANDING}$)	**41**	±12	**42**	±11.5	1.0	2.4	.852
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**26**	±6.8	**29**	±6	3.0	10.3	.138
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**53**	±8.6	**54**	±9.1	1.0	1.9	.667
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**26**	±5.7	**29**	±5.7	3.0	10.3	.089
$[RA]_{DROP}$ ($\%_{LANDING}$)	**71**	±12.8	**76**	±10.5	5.0	6.6	.197
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**12**	±5.7	**11**	±4.3	-1.0	-8.3	.713
DJ							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**52**	±6.1	**53**	±7.6	1.0	1.9	.821
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**25**	±7.8	**28**	±7.5	3.0	10.7	.224
$[ES]_{DROP}$ ($\%_{LANDING}$)	**41**	±9.3	**37**	±7.9	-4.0	-9.8	.171
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**29**	±5.3	**34**	±8.3	**5.0 #**	14.7	.033
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**51**	±8.1	**49**	±8.8	-2.0	-3.9	.437
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**29**	±7.3	**32**	±5.8	3.0	9.4	.075
$[RA]_{DROP}$ ($\%_{LANDING}$)	**73**	±13	**76**	±10.3	3.0	3.9	.346
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**14**	±7	**13**	±4.9	-1.0	-7.1	.711
DS							
$[EO]_{DROP}$ ($\%_{LANDING}$)	**52**	±6.9	**54**	±7.9	2.0	3.7	.396
$[EO]_{IMPACT}$ ($\%_{LANDING}$)	**22**	±6.4	**25**	±6.2	3.0	12.0	.185
$[ES]_{DROP}$ ($\%_{LANDING}$)	**37**	±9.9	**37**	±8.2	0.0	0.0	.912
$[ES]_{IMPACT}$ ($\%_{LANDING}$)	**29**	±6.7	**31**	±5.8	2.0	6.5	.318
$[TrA\text{-}IO]_{DROP}$ ($\%_{LANDING}$)	**53**	±8.2	**52**	±8	-1.0	-1.9	.576
$[TrA\text{-}IO]_{IMPACT}$ ($\%_{LANDING}$)	**27**	±6.4	**30**	±4.9	3.0	10.0	.103
$[RA]_{DROP}$ ($\%_{LANDING}$)	**73**	±12.5	**74**	±11.7	1.0	1.4	.606
$[RA]_{IMPACT}$ ($\%_{LANDING}$)	**12**	±5.6	**13**	±6.1	1.0	7.7	.552

sig. different from respective compared group ($p<0.05$)

Table A2.8: Descriptive (*Mean ± SD, Diff., Diff%*) and inferential comparison (*independent samples T-test*) of muscular *Phase* (*DROP, IMPACT*) activity (*%MVC*) between groups of *Familiarity* (*Familiar, Unfamiliar*)

	***Familiar** (N=25)*		***Unfamiliar** (N=18)*				
	Mean	*± SD*	*Mean*	*± SD*	*abs. Diff.*	*% Diff.*	*sig.p=*
DL							
$[EO]_{DROP}$ (%MVC)	**9**	±4	**12**	±6.4	**-3.5 #**	28.9	.034
$[EO]_{IMPACT}$ (%MVC)	**17**	±10.4	**22**	±10.3	-5.5	24.5	.097
$[ES]_{DROP}$ (%MVC)	**19**	±8.3	**24**	±9.9	-4.8	19.8	.091
$[ES]_{IMPACT}$ (%MVC)	**60**	±15.5	**68**	±22.5	-8.1	11.9	.170
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**39**	±14.3	**38**	±19.4	1.1	-2.7	.835
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**93**	±46.7	**86**	±39.6	7.3	-7.8	.593
$[RA]_{DROP}$ (%MVC)	**12**	±5.4	**13**	±6.2	-0.9	7.1	.613
$[RA]_{IMPACT}$ (%MVC)	**7**	±3.4	**8**	±3.7	-1.0	11.9	.377
DJ							
$[EO]_{DROP}$ (%MVC)	**8**	±3.9	**13**	±6.9	**-4.5 #**	35.1	.019
$[EO]_{IMPACT}$ (%MVC)	**20**	±12.6	**25**	±13.3	-4.9	19.7	.226
$[ES]_{DROP}$ (%MVC)	**23**	±8.1	**28**	±9.4	-5.0	17.6	.070
$[ES]_{IMPACT}$ (%MVC)	**80**	±27.3	**86**	±25.6	-6.2	7.3	.454
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**38**	±14.1	**38**	±14.9	0.4	-1.1	.924
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**110**	±65.4	**94**	±46.4	16.3	-14.8	.370
$[RA]_{DROP}$ (%MVC)	**14**	±7	**14**	±5.8	-0.3	2.1	.887
$[RA]_{IMPACT}$ (%MVC)	**9**	±4.1	**9**	±4.6	-0.6	5.9	.684
DS							
$[EO]_{DROP}$ (%MVC)	**8**	±3.8	**11**	±5.6	**-3.4 #**	30.0	.034
$[EO]_{IMPACT}$ (%MVC)	**16**	±10.3	**22**	±11.2	-5.9	26.7	.081
$[ES]_{DROP}$ (%MVC)	**18**	±5.8	**22**	±8.1	**-4.7 #**	21.1	.032
$[ES]_{IMPACT}$ (%MVC)	**67**	±18.7	**78**	±26	-10.7	13.9	.123
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**37**	±14.2	**34**	±13.2	2.8	-7.6	.514
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**95**	±58.6	**84**	±37.9	11.2	-11.8	.482
$[RA]_{DROP}$ (%MVC)	**11**	±5.2	**13**	±6.2	-2.0	14.9	.262
$[RA]_{IMPACT}$ (%MVC)	**7**	±3.5	**8**	±3.8	-0.9	11.0	.421

sig. different from respective compared group (p<0.05)

Table A2.9: Descriptive (*Mean ± SD, Diff., Diff%*) and inferential comparison (*independent samples T-test*) of muscular *Phase* (*DROP, IMPACT*) activity (*%MVC*) between groups of *Gender* (*Female, Male*)

	Females *(N=23)*		***Males*** *(N=20)*				
	Mean	*± SD*	*Mean*	*± SD*	*abs. Diff.*	*% Diff.*	*sig.p=*
DL							
$[EO]_{DROP}$ (%MVC)	**11**	±4	**9**	±5.6	-2	-18.2	.240
$[EO]_{IMPACT}$ (%MVC)	**19**	±8.4	**17**	±10	-2	-10.5	.504
$[ES]_{DROP}$ (%MVC)	**24**	±8.3	**18**	±7.8	**-6 #**	-25.0	.020
$[ES]_{IMPACT}$ (%MVC)	**69**	±19.4	**59**	±17	-10	-14.5	.106
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**41**	±14.8	**37**	±18.2	-4	-9.8	.397
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**93**	±46.6	**87**	±40.6	-6	-6.5	.646
$[RA]_{DROP}$ (%MVC)	**13**	±5.7	**12**	±5.8	-1	-7.7	.616
$[RA]_{IMPACT}$ (%MVC)	**8**	±4	**7**	±2.9	-1	-12.5	.300
DJ							
$[EO]_{DROP}$ (%MVC)	**11**	±5.3	**9**	±5	-2	-18.2	.124
$[EO]_{IMPACT}$ (%MVC)	**23**	±10.4	**21**	±11.6	-2	-8.7	.579
$[ES]_{DROP}$ (%MVC)	**29**	±9.1	**22**	±8.1	**-7 #**	-24.1	.014
$[ES]_{IMPACT}$ (%MVC)	**86**	±19.5	**84**	±29.8	-2	-2.3	.762
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**42**	±12.2	**34**	±15.5	-8	-19.0	.063
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**111**	±66	**95**	±48	-16	-14.4	.383
$[RA]_{DROP}$ (%MVC)	**14**	±6.7	**13**	±6.3	-1	-7.1	.662
$[RA]_{IMPACT}$ (%MVC)	**10**	±4.9	**8**	±3.4	-2	-20.0	.176
DS							
$[EO]_{DROP}$ (%MVC)	**10**	±4.1	**8**	±4.7	-2	-20.0	.188
$[EO]_{IMPACT}$ (%MVC)	**20**	±10.3	**17**	±9.6	-3	-15.0	.426
$[ES]_{DROP}$ (%MVC)	**22**	±7	**17**	±6.9	**-5 #**	-22.7	.038
$[ES]_{IMPACT}$ (%MVC)	**77**	±20.8	**66**	±20.5	-11	-14.3	.086
$[TrA\text{-}IO]_{DROP}$ (%MVC)	**39**	±12.3	**32**	±14.7	-7	-17.9	.106
$[TrA\text{-}IO]_{IMPACT}$ (%MVC)	**97**	±58.5	**83**	±40.2	-14	-14.4	.380
$[RA]_{DROP}$ (%MVC)	**13**	±5.6	**11**	±5.7	-2	-15.4	.337
$[RA]_{IMPACT}$ (%MVC)	**8**	±4	**7**	±3.2	-1	-12.5	.377

sig. different from respective compared group ($p<0.05$)

Table A2.10: *Spearman's rank correlation coefficient* (r_S) between muscular *PRE*-activity *(%MVC)* to $\alpha_{FLEX[LS\text{-}PV]}$ for cohort and groups of *Familiarity* (*Familiar, Unfamiliar*) and *Gender* (*Females, Males*)

	$\alpha_{FLEX[LS\text{-}PV]}$				
	Cohort	*Familiar*	*Unfamiliar*	*Females*	*Males*
n=	*43*	*25*	*18*	*23*	*20*
			DL		
CCI[EO|ES]	**.31***	.23	.15	-.03	**.72***
[EO]	**.39***	.28	.21	.14	**.68***
[ES]	.11	.07	-.04	.16	.01
[TrA-IO]	.10	.15	-.04	.02	.28
[RA]	.19	-.03	**.57***	.19	.13
			DJ		
CCI[EO|ES]	.21	.13	.12	-.07	**.53***
[EO]	.22	.12	.13	.03	**.45***
[ES]	.08	-.16	.32	.28	-.14
[TrA-IO]	-.03	-.03	.10	.10	.07
[RA]	.15	-.01	.39	-.10	.27
			DS		
CCI[EO|ES]	.27	.22	.15	-.14	**.71***
[EO]	**.31***	.32	.05	-.00	**.65***
[ES]	.15	.18	-.12	.37	-.10
[TrA-IO]	.03	-.01	.07	-.07	.14
[RA]	.25	.19	.14	.17	.37

***** sig. rank association ($p<0.05$)

APPENDIX 2

***Table A2.11:** Spearman's rank correlation coefficient* (r_S) between muscular activity (*%MVC*) during *DROP* and *IMPACT* to $\alpha_{FLEX[LS\text{-}PV]}$ for cohort and groups of *Familiarity* (*Familiar, Unfamiliar*) and *Gender* (*Females, Males*)

		$\alpha_{FLEX[LS\text{-}PV]}$				
		Cohort	Familiar	Unfam.	Females	Males
	n=	*43*	*25*	*18*	*23*	*20*
		DL				
CCI[EO/ES]	*DROP*	.27	.23	.01	.03	**.55***
	IMPACT	**.35***	.28	.2	.05	**.78***
[EO]	*DROP*	**.31***	.23	.08	.14	**.49***
	IMPACT	**.36***	.26	.26	.08	**.71***
[ES]	*DROP*	**.30***	.21	.09	.31	.25
	IMPACT	.11	.08	-.05	.27	-.01
[TrA-IO]	*DROP*	.08	.10	-.10	.18	.06
	IMPACT	.11	.17	.01	.02	.30
[RA]	*DROP*	.02	-.19	.29	.03	-.01
	IMPACT	**.34***	.14	**.64***	.33	.32
		DJ				
CCI[EO/ES]	*DROP*	.26	.25	-.04	.02	**.53***
	IMPACT	.20	.08	.20	-.08	**.51***
[EO]	*DROP*	**.34***	.37	.05	.23	**.49***
	IMPACT	.18	.10	.16	-.04	.44
[ES]	*DROP*	.19	-.07	.35	.32	.06
	IMPACT	.08	-.14	.26	.36	-.22
[TrA-IO]	*DROP*	.04	.08	-.14	.04	.07
	IMPACT	-.05	-.11	.17	-.14	.08
[RA]	*DROP*	-.04	-.04	-.15	-.16	.18
	IMPACT	.28	.00	**.65***	**.42***	.08
		DS				
CCI[EO/ES]	*DROP*	.19	.14	.07	-.10	**.49***
	IMPACT	.26	.22	.12	-.16	**.74***
[EO]	*DROP*	.26	.29	-.08	.08	.42
	IMPACT	**.33***	.33	.12	.02	**.68***
[ES]	*DROP*	**.30***	.39	-.07	.28	.33
	IMPACT	.09	.11	-.09	.31	-.14
[TrA-IO]	*DROP*	.04	.08	-.10	.00	.10
	IMPACT	.03	-.01	.10	-.07	.16
[RA]	*DROP*	.03	-.04	-.13	-.05	.21
	IMPACT	**.33***	.21	.43	.35	.28

* sig. rank association (p<0.05)

APPENDIX 2

Table A2.12: *Spearman's rank correlation coefficient* (r_S) between muscular contribution ($\%_{LANDING}$) during *DROP* and *IMPACT* to $\alpha_{FLEX[LS-PV]}$ for cohort and groups of *Familiarity* (*Familiar, Unfamiliar*) and *Gender* (*Females, Males*)

		$\alpha_{FLEX[LS-PV]}$				
		Cohort	Familiar	Unfam.	Females	Males
	n=	*43*	*25*	*18*	*23*	*20*
				DL		
[EO]	*DROP*	.02	-.08	.04	.25	-.17
	IMPACT	.10	.25	-.10	.06	.20
[ES]	*DROP*	.12	.02	.16	.10	.19
	IMPACT	.14	.29	.04	.17	.17
[TrA-IO]	*DROP*	.04	.03	.31	.22	-.16
	IMPACT	.14	.31	-.06	.15	.20
[RA]	*DROP*	-.05	.03	-.16	-.07	-.05
	IMPACT	.17	.25	.29	.18	.13
				DJ		
[EO]	*DROP*	.09	.17	-.12	.20	-.02
	IMPACT	-.08	.01	-.14	-.19	-.03
[ES]	*DROP*	.09	.04	.17	-.04	.26
	IMPACT	-.21	-.16	-.29	-.11	-.42
[TrA-IO]	*DROP*	.19	.21	.31	.27	-.03
	IMPACT	-.23	-.17	-.21	-.21	-.27
[RA]	*DROP*	-.24	-.15	-.37	**-.42***	-.02
	IMPACT	.15	.15	.25	.32	-.07
				DS		
[EO]	*DROP*	-.06	-.05	-.17	.13	-.27
	IMPACT	.12	.16	-.10	-.14	.41
[ES]	*DROP*	.05	.16	-.11	-.18	.36
	IMPACT	-.07	-.05	-.34	-.02	-.18
[TrA-IO]	*DROP*	.12	.26	.13	.24	-.06
	IMPACT	-.03	.04	-.22	-.16	.14
[RA]	*DROP*	-.18	-.11	-.20	-.35	.03
	IMPACT	.18	.20	.15	.30	.05

* sig. rank association (p<0.05)

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