The Biomechanical Demands of Ballet – Implications for Performance Profiling and Injury Aetiology

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Abstract

Ballet epidemiology audits have established a high prevalence of ankle injury. However, aetiological research specific to these injuries in a ballet context is limited. Given the mechanical complexity of ballet, the aim of the current thesis was to conduct a multi-modal biomechanical investigation of ballet-specific movement. In study 1, amateur female ballet dancers completed a testing battery containing the following seven ballet-specific jump landing tasks; jeté, jeté step, échappé, sissonne, sissonne pas de bourres (PDB), temps levé and jeté en tournant (ET). A multi-camera automated motion capture system with integrated, synchronous force plate analysis was used to quantify kinetic and ankle joint kinematic responses to the movement battery. Resultant biomechanical responses were task-dependent, and bilateral symmetry was evident across all movements. Study 2 quantified the electromyographic responses to the same movement battery, again demonstrating bilateral symmetry, and mirroring the hierarchical ordering of movement demand demonstrated in study 1. Study 3 evaluated bilateral isokinetic ankle strength (Peak Torque, Angle of Peak Torque, Functional Range, Angle-Specific Torque, Dynamic Control Ratios) using a dynamometer (Biodex System 4 pro), with a testing protocol supported by the joint angular velocity and displacement data obtained during study 1. Ballet dancers are eccentric inversion strength dominant, exacerbated at $\geq 60°\cdot s^{-1}$, and demonstrate lateral symmetry in strength which may reflect a chronic training adaptation.

Study 4 utilised accelerometry to quantify planar mechanical load response to the movement battery, consistently identifying the task-dependent hierarchy of movement response, and demonstrating strong correlations ($r \geq 0.59$) with planar ground reaction force measures in study 5. Study 6 utilised the greater ecological validity afforded by accelerometry to investigate the mechanical implications of consecutive ballet performances. There was some evidence of a fatigue response between bouts, and anatomical location of the accelerometer device was an important consideration when interpreting movement response. The studies comprising this thesis offer a contemporary approach to conducting biomechanical analyses in ballet. Inclusion of a ballet-specific movement battery offers a novel dimension and increases ecological validity. Synergy and cohesion within a multi-modal design was reflected in the responses across data collection tools, demonstrating innovation in ballet research. The findings provide important clinical information which may help develop the understanding of injury occurrence. Utilising laboratory- and field-based monitoring techniques in a ballet context may better inform clinical practice in training prescription towards reducing injury.

Key Words: Accelerometry, Aetiology, Ballet, Biomechanics, Epidemiology, Female Dancers, Injury.

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List of Abbreviations, Acronyms and Symbols

ACC_{AP}	Anteroposterior Accelerations
ACC_{ML}	Mediolateral Accelerations
ACC_{V}	Vertical Accelerations
ACL	Anterior Cruciate Ligament
AIM	Automatic Identification of Markers
ANOVA	Analysis of Variance
APT	Angle of Peak Torque
AST	Angle-specific Torque
BW	Body Weight
CAI	Chronic Ankle Instability
CAIT	Cumberland Ankle Instability Tool
CI	Confidence Intervals
CON_{EV}	Concentric Eversion
CON_{INV}	Concentric Inversion
COP	Centre of Pressure
D	Dominant
DAFT	Dance Aerobic Fitness Test
DCR	Dynamic Control Ratio
DCR_{AST}	Dynamic Control Ratio derived from Angle-Specific Torque
DCR_{PT}	Dynamic Control Ration derived from Peak Torque
DL	Dominant Limb
ECC_{INV}	Eccentric Inversion
EMG	Electromyography
ET	En Tournant
FR	Functional Range
GPS	Global Positioning System
GRF	Ground Reaction Force
HR	Heart Rate
Hz	Hertz
IC	Initial Contact
iEMG	Integrated EMG
LG	Lateral Gastrocnemius
MVIC	Maximal Voluntary Isometric Contraction

ND	Non-dominant
NDL	Non-dominant Limb
PDB	Pas de Bourres
PL	Peroneus Longus
PL_{AP}	Anteroposterior PlayerLoad™
$PL_{AP\%}$	Anteroposterior Contribution to PL_{TOTAL}
PL_{ML}	Mediolateral PlayerLoad™
$PL_{ML\%}$	Mediolateral Contribution to PL_{TOTAL}
PL_{TOTAL}	Accumulated PlayerLoad™
PL_{V}	Vertical PlayerLoad™
$PL_{V\%}$	Vertical Contribution to PL_{TOTAL}
PT	Peak Torque
QTM	Qualisys Track Manager
RMS	Root Mean Square
RPE	Rating of Perceived Exertion
$SAFT_{90}$	90-minute Soccer Aerobic Fitness Test
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
TA	Tibialis Anterior
TO	Toe Off
$_{V}ACC$	Vertical Accelerations
$_{V}GRF$	Vertical Ground Reaction Force
$_{V}MLR$	Vertical Mean Loading Rate

Chapter 1. Introduction

1.1 Overview

Amongst the many complex and technical definitions of sports biomechanics observed in the literature, perhaps 'the study and analysis of human movement patterns in sport' is the simplest, yet most appropriate (Bartlett, 2007). Sports biomechanics is a research discipline with two inter-dependent branches; injury prevention/reduction and performance enhancement (Yeadon and Challis, 1994). Irrespective of the research focus, experimental approaches aim to address the What? How? and Why? questions posed by the execution of sport, providing valuable information to research professionals, coaches and athletes alike (Yeadon and Challis, 1994). Data on the locomotor profile of the sport, and the athlete, may highlight elements of a movement associated with skilled performance, whilst identifying markers that proliferate the risk of injury.

Ballet is primarily an art form, with performance excellence largely dependent on artistic grace and merit during choreographed routines (Hincapie, Morton and Cassidy, 2008). However, ballet dictates a high level of athleticism owing to the many complex techniques comprising its movement profile (Russell, 2013). Akin to other sports modalities, participation in ballet has an inherent risk of injury. The epidemiology of ballet injuries is well defined, in that overuse traumas are more common, and the foot and ankle complex represents the prime injury location (Smith et al., 2015). However, comparatively less is understood from an aetiology perspective, but the prevalence of these injuries may be attributed to the high volume of mechanically demanding jump landing tasks performed in training and competition (Twitchett, Koutedakis and Wyon, 2009). Given the limited data available, further biomechanical investigations of the potential mechanisms and associated risk factors for injury are warranted.

Research investigating the biomechanics of ballet is a burgeoning line of enquiry, and contemporary advancements in technology have broadened the array and sophistication of relevant analysis tools towards improved rigour. Despite this, existing methodological designs have some noteworthy limitations which may hinder understanding on the characteristics of ballet movement, and, potential implications for injury occurrence. For example, previous kinematic and kinetic assessments of jump landings in ballet have used clinically-focused drop/jump landing tasks (Orishimo et al., 2009; Liederbach et al., 2014), or, a solitary ballet-specific manoeuvre (Kulig, Fietzer and Popovich, 2011; Peng et al., 2015). Thus, relevant studies often lack specificity, negating many of the jump landing tasks allied to ballet performance. With 3D motion capture and force platform analyses dominating pertinent literature, Electromyography (EMG) and Isokinetic Dynamometers are used less frequently. In the relatively few studies available, EMG has been deployed predominantly to quantify the neuromuscular contributors to discrete floor tasks (Lin et al., 2014; Tanabe, Fujii and Kouzaki, 2017; Zaferiou et al., 2017), whilst isokinetic strength protocols have seemingly been designed without consideration of injury epidemiology (Koutedakis and Sharp, 2004; Tsanaka, Manou and Kellis, 2017; Lima et al., 2018). Currently, there is an evident lack of research adopting a multi-modal approach to investigate the biomechanics of ballet-specific movements. For research to progress in this domain, flexibility in the synthesis and cohesion of multiple analysis tools may provide the most comprehensive profiling of ballet motion. That is, a kinematic and kinetic assessment of a battery of ballet-specific tasks, designed to comprise both uni- and bilateral landings, with a consideration of linear and rotational demand, and a differentiation between hold and transitional landings, but supplemented by a synchronous consideration of the neuromuscular strategies governing movement. In addition, an evaluation of isokinetic strength to quantify mechanical capacity in dancers, but with a protocol supported by joint angular velocity and displacement data from kinematic analyses of ballet-specific movements. Though restricted to laboratory settings, this innovation and novelty in methodological design heightens both ecological validity, and the level of biomechanical understanding regarding ballet movement.

Despite the importance of acquiring kinematic and kinetic information, opportunities to enhance current ballet performance monitoring strategies may be achieved via field-based biomechanical assessments. The invention of portable, wearable micro-technologies has resulted in a surge of research investigating performance rigours both in training and competition (Malone et al., 2017). Global Positioning Systems (GPS) devices often house inertial sensor components such as tri-axial accelerometers, enabling the assessment of the loads imposed by sport to be conducted beyond laboratory conditions (Waldron et al., 2011; Hausler, Halaki and Orr, 2016). Accelerometers can be worn during performance to quantify the movement characteristics of ballet, and the workload tolerances of dancers to which there is limited current information (Brogden et al., 2018). Recent evidence has demonstrated positive associations between accelerometry and conventional measures of load during jump landing tasks (Simons and Bradshaw, 2016; Setuain et al., 2016). Hence, there is potential for the same observations in ballet, meaning biomechanics research can be conducted in real-word settings which contrasts with the majority of existing studies. The findings from the current thesis may add to the injury aetiology knowledge in ballet, whilst the methodological constructs may enhance current practice towards athlete monitoring.

1.2 Aims and Structure of Thesis

The overall aim of the current thesis was to conduct an extensive biomechanical assessment of ballet-specific movement using a multi-modal approach. The findings from the thesis may develop understanding on the occurrence of injuries, and enhance current ballet monitoring practices towards reducing and managing injury risk. Specific thesis objectives relating to the aim were to:

- Examine bilateral 3D kinematic and external kinetic responses to a battery of ballet-specific jump landing tasks.
- Quantify the neuromuscular responses of lower limb musculature to the selected tasks.
- Evaluate ankle eversion/inversion strength using a functionally relevant isokinetic strength testing protocol.
- Assess the multi-planar 'load' profiles of the ballet-specific movements using tri-axial accelerometry.
- Explore potential associations between accelerometry and kinetic metrics.
- Investigate the utility of accelerometry to quantify acute and cumulative load responses to a choreographed ballet routine.

The thesis presents six discrete experimental studies, designed to reflect the range of data collection tools required for a comprehensive analysis of ballet, and the transition from laboratory to field-based assessments. Following a thorough review of ballet epidemiology and injury aetiology literature, a General Methodology chapter outlines information relating to the participant recruitment process. In addition, research design considerations in developing a battery of ballet-specific movements and a choreographed routine, and a methodological philosophy of synthesis between the biomechanical analysis tools is contained within. The initial investigations of the thesis (studies 1-4) were conducted in the laboratory. Experimental study 1 comprises a 3D kinematic and kinetic evaluation of a range of ballet-specific jump landing tasks using a multi-camera, automated motion capture system with synchronous force plate analysis. Joint kinematics are focused on the ankle in accordance with the epidemiology of ballet injuries. EMG is used in experimental study 2 to quantify muscular responses to the movement battery completed in the preceding chapter. Surface EMG is applied to the Peroneus Longus, Tibialis Anterior, and Lateral Gastrocnemius muscles given their functional responsibilities. Experimental study 3 presents an isokinetic dynamometry assessment of the same dancer cohort used in the previous studies. Ankle eversion and inversion strength was examined, reflecting the common mechanism of injury. Kinematic data relating to ankle joint displacements and angular velocities obtained in study 1 supported the testing protocol. EMG responses to maximal strength testing were also measured, and were compared with those obtained during the movement battery in study 2. The relative portability for in-vivo assessments of ballet offered by EMG in contrast to motion analysis, force platform analysis and isokinetic dynamometry, is developed in experimental study 4 with an evaluation of tri-axial accelerometry to the same battery of movements. Absolute and relative PlayerLoad™ metrics are compared between movements to distinguish the planar characteristics specific to each task. Associations between accelerometry and ground reaction force analysis are then explored in experimental study 5, with a particular focus on the utility of accelerometry for valid biomechanical assessments in field-based designs. Experimental study 6 further develops the field-based research paradigm, utilising tri-axial accelerometry to quantify the mechanical responses to a choreographed ballet routine. Specifically, a multi-stage performance piece with progressive intensity, constructed using the battery of tasks completed in study 1.

The current thesis therefore aims to develop a comprehensive biomechanical understanding of ballet-specific movement. The logical structure and design of each study demonstrates cohesion between investigations, and offers rigour in the assessment of ballet, and the functional capacities of dancers. Synergy within a multi-modal design strengthens the level of analysis, and the development of an ecologically valid approach culminates in a biomechanical analysis of ballet performance in the last experimental study. A concluding synopsis considers the thesis in relation to the initial aims and objectives. The key study findings are discussed, and the practical implications of the body of research for both ballet practitioners and biomechanics researchers are outlined.

Chapter 2. Review of Literature

The aim of the thesis is to conduct a multi-modal biomechanical investigation of ballet-specific demands, with implications for performance profiling and injury aetiology. The literature review initially considers injury epidemiology in ballet populations, with specific focus on the prevalence of injury, and the primary anatomical location for injury. The subsequent section considers the multi-factorial aetiology of injury to inform the development of a comprehensive biomechanical evaluation, and, the mechanism common to injury to inform the application of the biomechanical analyses. A critical evaluation on the use of biomechanical analysis tools in existing ballet research concludes the review, with specific focus on studies pertinent to the current thesis.

2.1 Epidemiology of Ballet Injury

Injury epidemiology research is conducted to inform the strategies aimed at reducing injury in order to keep performers in training and competition (Langeveld, Coetzee and Holtzhauxen, 2012). Despite advances in screening protocols, and the identification of athletes a greater risk, information in Table 2.1 highlights that irrespective of genre, dancers are highly susceptible to injury (Cho et al., 2009; Yau et al., 2017). Whilst existing data emphasises a high occurrence of injury in dance, the contrasting figures are likely influenced by the locomotor characteristics comprising each genre (Krasnow et al., 2011), and, a lack of uniformity in capturing workloads and defining injury (Hincapie, Morton and Cassidy, 2008; Jacobs, Hincapie and Cassidy, 2012; Jacobs et al., 2017). The contentious issues surrounding injury quantification have been scrutinised by the Standards Measure Consensus Initiative of the International Association for Dance Medicine and Science (Liederbach et al., 2012). Therefore, caution should be applied when examining injury epidemiology literature, as meaningful comparisons may only be achieved across studies with parallel methodological designs and injury definitions (Russell, 2013).

Table 2.1. The prevalence of injury across various dance genres.

Author	Genre	Number of Subjects	Injured Dancers (% of Sample)	Number of injuries
Yau et al., 2017	Ballet & Contemporary	480	362 (75)	1084
Cahalan et al., 2016	Irish Dance	85	70 (82)	278
Ojofeitimi, Bronner and Woo, 2012	Hip Hop	312	232 (74)	738
Ojofeitimi and Bronner, 2011	Modern	87	75 (86)	217
Cho et al., 2009	Break	42	40 (95)	193

In acknowledgment of ballet being one of most popular dance genres in terms of participation rates (Malkogeorgos et al., 2011), and that many studies contained in injury surveillances have focused on this population (Smith et al., 2015), the epidemiology of injuries in ballet will be discussed hereafter. Injury data is categorised into frequency, nature, and anatomical location.

Frequency of Ballet Injuries

The data presented in Table 2.2 relates to the occurrence of injury in ballet populations. The information highlights that musculoskeletal disorders are prevalent in 13%-100% of ballet dancers (Negus, Hopper and Briffa, 2005; Comin et al., 2013). A further observation demonstrates a tendency towards a higher prevalence (42%-100%) in amateur dancers (Gamboa et al., 2008; Negus, Hopper and Briffa, 2005) compared with their professional counterparts (13%-75%, Comin et al., 2013, Byhring and Bo, 2002).

Injury statistics displayed in Table 2.3 indicate that injury incidence rates range between 0.4-4.8 per 1000 hours of dance exposure in the amateur and professional cohorts (Leanderson et al., 2011; Allen et al., 2012). Further extrapolation demonstrates that injury incidences attributed to gender and training status are comparable (Smith et al., 2015), with female and male amateur dancer sustaining 1.77 and 2.12 injuries per 1000 dance hours, compared with 1.06 and 1.46 in their professional counterparts. Overall incidences of 1.42 and 1.79 are reported for females and males respectively.

Table 2.2. The prevalence of injuries in ballet dancers. Adopted and modified from Smith et al., (2015).

Author	Population	Age (years) Mean ± SD	Number of Subjects (Female/Male)	Injured Dancers (% of sample)	Number of Injuries
Sorbrino and Guillen, 2017	Amateur	Junior (≤ 21) Intermediate (22-31) Senior (≥ 32)	145 (Junior 49: Intermediate 74: Senior 22).	Not Specified	486
Caine et al., 2016	Amateur	11-21	71 (44/27)	61 (86)	114
Ramkumar et al., 2016	Professional	27.5	153 (81/72)	Not Specified	574
Bowerman et al., 2014	Amateur	16 ± 1.58	46 (30/16)	29 (63)	59
Ekegren, Quested and Brodrick, 2014	Amateur	17.2 ± 1.21	266 (154/112)	203; 117 F, 86 M (76)	378
Wyon et al., 2014	Professional	26 ± 4.57	24 (13/11)	12 (50)	12
Allen et al., 2013	Professional	Season 1: 25 ± 5 F, 23 ± 4 M. Season 2: 25 ± 5 F, 24 ± 4 M Season 3: 26 ± 5 F, 24 ± 4 M	Season 1: 52 (27/25) Season 2: 58 (29/ 9) Season 3: 53 (27/26)	Not Specified	Season 1: 355 Season 2: 183 Season 3: 174
Comin et al., 2013	Professional	26.4 ± 6.6 F, 28.7 ± 5.8 M	79 (44/35)	10 (13)	14
Leanderson et al., 2011	Ballet Dancers	10-21	476 (297/179)	210 (44)	438

Table 2.2. *Continued*.

Campoy et al., 2011	Not Specified	18.3 ± 4.7	500 (409, 91)	Not Specified	377
Gamboa et al., 2008	Amateur	14.7 ± 1.9	359 (228, 71)	151 (42)	198
Liederbach, Dilgen and Rose, 2008	Mixed	Not Specified	117 (64, 53)	Not Specified	1427
Negus, Hopper and Briffa, 2005	Amateur	18	29	29 (100)	82
Byhring and Bo, 2002	Professional	26 ± 5.7 F, 27 ± 4.6 M	41 (27/14)	31 (75)	64
Coplan, 2002	Amateur	22	30 (27/3)	14 (47)	22
Nilsson et al., 2001	Professional	28.3	78 (46/32)	98	390

F, Female; M, Male.

Table 2.3. Injury incidences among amateur and professional ballet dancers. Adopted and modified from Smith et al., (2015).

Study	Study Period	Dancers, n (Female/Male)	Injured Dancers, n	Total Injuries, N	Total Hours	Injury Incidence (per 1000 dance hours) Female	Male	Overall
Amateur								
Caine et al., 2016	1 season	71 (44/27)	61	114	50,346	2.73	3.48	3.06
Bowerman et al., 2014	6 months	46 (30/16)	29	59	24,583	2.19	2.81	2.4
Ekergren et al., 2014	1 school year	266 (154/112)	203	378	274,089	1.36	1.38	1.38
Leanderson et al., 2011	7 school years	476 (297/179)	210	438	555,318	0.8	0.8	0.79
					35,333	0.3	0.5	0.37
					258,571	0.7	0.6	0.66
					261,414	0.9	1.1	0.97
Gamboa et al., 2008	5 school years	359; mean, 71.8 per year (288/71)	151	198	257,143	-	-	0.77
Total		1218 (813/405)	654	1187	1,161,479	1.77	2.12	1.68
Professional								
Wyon et al., 2014	4 months	24 (13/11)	13	12	12,768	1.01	0.85	0.94
Nilsson et al., 2001	5 seasons	390 (average 78 per season)	98	390	629,032	0.56	0.70	0.62
Total		577 (96/91)	111	402	896,942	1.06	1.46	1.24
Overall Injury Incidence				1,589	2,058,421	1.42	1.79	1.46

n, number.

Nature of Ballet Injuries

The information presented in Table 2.4 shows that overuse type traumas account for a greater proportion of injuries in the amateur populations (73.4%) compared with professional dancers (57%). Further, there is a lack of gender disparity in overuse injury rates in the amateur population (females, 74.6%; males, 70.1%), but a greater frequency of overuse trauma is evident in professional female dancers (64.4%) compared with their male equivalents (50.1%).

Table 2.4. Nature of injures among amateur and professional ballet dancers. Adopted and modified from Smith et al., (2015).

			Overall Injuries, n (%)		Female Injuries, n (%)			Male Injuries, n (%)		
Study	**Dancers, n**	**Injuries, n**	**Traumatic**	**Overuse**	**Traumatic**	**Overuse**	**Total**	**Traumatic**	**Overuse**	**Total**
Amateur										
Sorbrino and Guillen, 2017	145	486	120 (24.7)	366 (75.3)	62 (24.1)	195 (75.9)	257	58 (25.3)	171 (74.7)	229
Yau et al., 2017	480	1014	325 (32.1)	689 (67.9)	222 (29.1)	542 (70.9)	774	103 (41.2)	147 (58.8)	250
Caine et al., 2016	71	114	39 (34.2)	75 (65.8)	-	-	-	-	-	-
Ekegren, Quested and Brodrick, 2014	266	378	106 (28)	272 (72)	-	-	-	-	-	-
Leanderson et al., 2011	476	438	101 (23.1)	337 (76.9)	62 (23)	207 (77)	269	39 (23.1)	130 (76.9)	169
Gamboa et al., 2008	204	198	54 (27.3)	144 (72.7)	-	-	-	-	-	-
Negus, Hopper and Briffa, 2005	29	82	14 (17.1)	68 (82.9)	-	-	-	-	-	-
Total	1,680	2,710	759 (26.6)	1951 (73.4)	346 (25.4)	944 (74.6)	1,290	200 (29.9)	448 (70.1)	648
Professional										
Allen et al., 2013	163	712	308 (43.3)	404 (56.7)	109 (33.7)	214 (66.3)	323	199 (51.2)	190 (48.8)	389
Nilsson et al., 2001	98	390	166 (42.6)	224 (57.4)	80 (38.5)	128 (61.5)	208	86 (47.3)	96 (52.7)	182
Total	261	1102	474 (43)	628 (57)	189 (35.6)	342 (64.4)	531	285 (49.9)	286 (50.1)	571
Overall	1941	3812	1233 (32.3)	2579 (67.7)	535 (29.4)	1286 (70.6)	1,821	485(40)	734 (60)	1219

Anatomical Location of Ballet Injuries

Of all injuries sustained in ballet, between 59%-93% occur in the lower extremities (Bowerman et al., 2014; Ramkumar et al., 2016). Observations from the data displayed in Table 2.5 show that the most common injury location in amateur and professional ballet dancers is the ankle and foot (14%-57%; Byhring and Bo, 2002; Liederbach, Dilgen and Rose, 2008), followed by the knee (7.3%-36%; Coplan, 2002; Negus, Hopper and Briffa, 2005), hip/groin (4.5%-25.6%; Coplan, 2002; Negus, Hopper and Briffa, 2005), and thigh (1.2%-6%; Negus, Hopper and Briffa, 2005; Sorbrino and Guillen, 2017).

Summary

Despite the varying rates of injury occurrence in ballet, existing studies have unequivocally established that ballet presents a high risk of injury to performers (Smith et al., 2015; Caine et al., 2016; Jacobs et al., 2017). With overuse type pathologies representing a large proportion in the amateur female cohort (74.6%), and that injuries are primarily localised to the ankle complex, biomechanics research with a consideration of these observations is a necessary and worthwhile endeavour.

Table 2.5. Distribution of injuries in amateur, professional, and unspecified/mixed ballet dancers. Adopted and modified from Smith et al., (2015).

		Injury Location		
Study	**Injuries, n**	**Head/Upper Extremity/Back**	**Lower Extremity**	**Foot and Ankle**
Amateur				
Yau et al., 2017	1014	19%, (head and neck, 2%; back, 14%; trunk or abdomen, 1%; shoulder, 2%)	81% (hip or thigh, 15%; knee, 13%; lower leg, 7%)	44% (foot or toe, 20%; ankle 25%)
Sorbrino and Guillen, 2017	366	23% (upper limbs, 0.3%; shoulder, 2.2%; spine, 20.5%)	77% (hip and pelvis, 14.5%; thigh, 6%; knee, 18%; leg, 3.8%)	34.7% (ankle, 19.9%; foot, 14.8%)
Caine et al., 2016	114	14% (head/spine trunk, 8%; neck, 3%; lower back, 3%)	86% (hip/groin, 19%; thigh, 3%; knee, 15%; lower leg (shin and calf, 10%)	39% (Achilles tendon, 8%; ankle, 15%; foot, 9%; toes, 7%)
Bowerman et al., 2014	59	Lumbar Spine, 7%	93% (hip, 24%; upper leg, 7%; lower leg, 9%)	46% (ankle, 15%; foot, 31%)
Ekegren, Quested and Brodrick, 2014	378	Head and neck, 3%; UE, 3% (shoulder, 64% of UE); trunk, 16% (lumbar spine, 60% of trunk)	77% (hip/groin, 7%; thigh, 2%; knee, 10%; lower leg (shin and calf), 17%)	39% (ankle, 35%; foot, 15%; unspecified, 3%).
Leanderson et al., 2011	438	UE/misc, 3.9%; back, 13%	83.1% (hip/thigh, 11%; knee, 21%; lower leg/foot, 52%)	-
Gamboa et al., 2008	198	Back, 9%	91% (hip, 22%; knee, 16%)	53% (all ankle and foot)
Negus, Hopper and Briffa, 2005	82	Low back, 9.8%	90.2% (hip, 25.6%; thigh, 1.2%; knee, 7.3%; lower leg, 19.5%)	36.6% (ankle, 25.6%; foot, 11%)
Coplan, 2002	22	Low back, 13.6%	86.4% (hip, 4.5%; knee, 36%; shin, 22.7%)	18.1% (ankle, 13.6%; foot, 4.5%)

Table 2.5. *Continued.*

Professional				
Ramkumar et al., 2016	574	41% (shoulder, 2%; cervical spine, 9%; thoracic spine, 7%; lumbar spine, 20%, elbow and hand/wrist, 3%; other, 2%)	59% (hip, 5%; knee 8%; leg 4%)	38% (all ankle and foot)
Byhring and Bo, 2002	64	Neck, 9%; shoulder/arm, 4%; upper back, 11%; low back, 8%	68% (hip, 14%; knee, 16%; leg, 24%)	14% (ankle, 3%; foot, 11%)
Nilsson et al., 2001	390	UE, 7.2%; lower back/gluteal region, 17.9%; misc, 1.9%	71.6% (thigh/groin, 4%; knee, 11%; lower leg, 3%)	54% (all foot and ankle)
Unspecified/Mixed				
Campoy et al., 2011 (Unspecified)	320	Upper limbs, 14.37%; trunk/hip, 7.19%	78.44% (thigh/leg, 27.5%; knee, 22.19%)	28.75% (all foot and ankle)
Liederbach, Dilgen and Rose, 2008 (Mixed)	1427	UE, 3%; spine, 12%; other, 13%	72% (hip, 9%; knee, 6%)	57% (all foot and ankle)

n, number; misc, miscellaneous.

2.2 Aetiology of Ankle Injury in Ballet

The ankle joint is the kinetic junction between structures of the lower leg and foot, enabling the body to move. The ankle complex comprises a network of bones, muscles and ligaments, which together are responsible for managing the resultant compressive and shear forces produced during gait (Michael et al., 2008). Hence, stability and optimal function within the joint and surrounding tissues is integral to mobilisation, and, a reduced risk of injury (Brockett and Chapman, 2016). The ankle joint functions as a multi-axis structure; the foot is able to plantar- and dorsiflex in the sagittal plane, internally/externally rotate in the transverse plane, and evert/invert in the coronal plane. Movement which occurs beyond the physiological motion capabilities of the joint, provides mechanical and injury risk implications for the bony and soft tissues.

The aetiology of ankle injuries in ballet dancers towards subsequent biomechanical investigation requires a knowledge and understanding of the primary mechanisms of injury, and, the associated risk factors (Bahr and Krosshaug, 2005). In accordance with Meeuwisse', (1994) dynamic model (Figure 2.1), it is vital to consider the multi-factorial nature of injury causation, owing to the complex interrelationships between the mechanism and the risk factors.

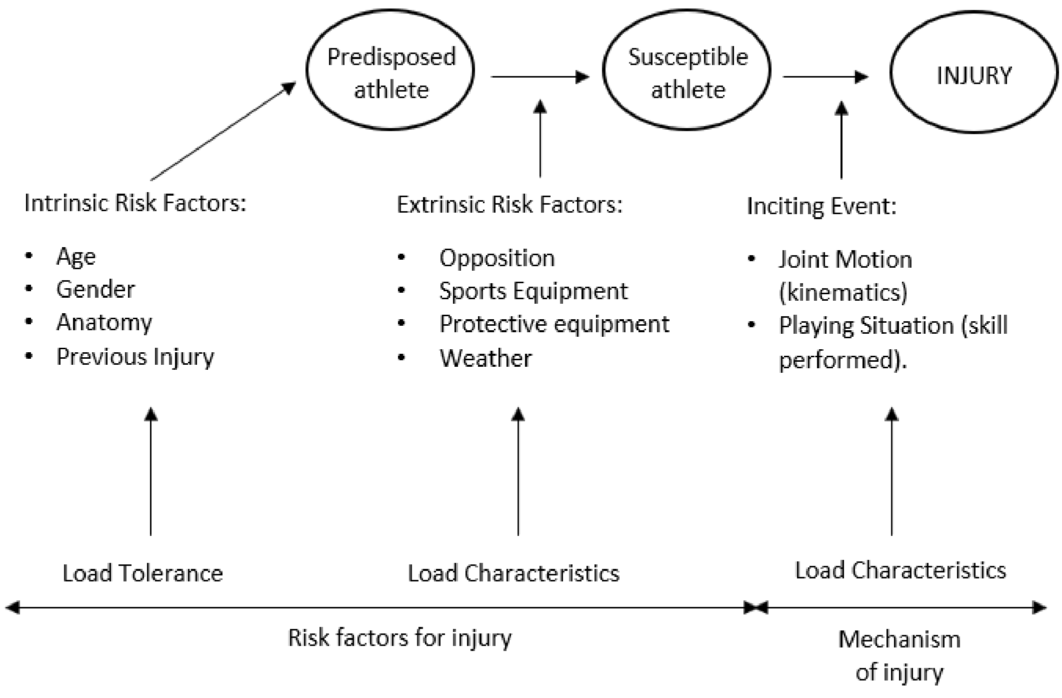

Figure 2.1. A dynamic model of the interrelationships between the risk factors and mechanisms for sport injury – recreated from Meeuwisse, (1994).

Mechanism of Injury

The inciting event, whether a contact or non-contact mechanism, either produces an intolerable load for the tissues, or, reduces the tolerance levels to a point where normal load causes mechanical failure and resultant injury (Meeuwisse et al., 2007). Non-contact injury cases are arguably more concerning, as they align closely with both the movement profile of a chosen sport, and the physical conditioning/capacity of the athlete. Moreover, sport science interventions via appropriate regulation of training loads and recovery, can help to mitigate against non-contact injuries (Eckard et al., 2018). Sports injury clinicians widely agree that jump landing techniques and manoeuvres that require rapid acceleration and/or deceleration with a change of direction (i.e. a cut), present as the primary locomotor contributors to ankle injury (Fousekis, Tsepis and Vagenas, 2012). Execution of these particular movements orientate the foot into plantarflexion and eversion/inversion, placing stress on the bony and ligamentous structures of the ankle complex (Funk, 2011). Exposure to these joint displacements during repetitive, high impact ground strikes proliferates injury potential, which may explain the high prevalence of ankle injury in the ballet dance populations (Moita et al., 2017). Given the aesthetic importance of ballet performance, discrete technical elements of a routine dictate extreme ankle joint positions such as the pointe technique (O'Laughlin, Hodgkins and Kennedy, 2008), which exacerbates the risk of injury to dancers. Knowledge surrounding the multi-faceted nature of ankle injury risk factors may provide vital clinical information to inform the strategies deployed in managing injury risk.

Risk Factors for Injury

A risk factor for sport injury is any attribute or characteristic of sport and exercise performance that increases the risk of sustaining an injury (Hopkins et al., 2007). Injury risk factors do not infer cause, rather, athletes who present risk factors are statistically more likely to sustain an injury compared to athletes without risk factors (Windt and Gabbett, 2016). Risk factors are typically divided into two main categories: intrinsic factors defined as those relating to the athlete, and extrinsic factors, those relating to the sport and exercise environment. Risk factors can be further partitioned into modifiable, distinguished as factors that can be manipulated towards a reduced injury risk, and non-modifiable which are out of human control (Emery and Tyreman, 2009). Knowledge on non-modifiable risk factors such as gender and age is important to identify athletes at an increased risk, but perhaps modifiable factors are of greater interest to professionals involved in mitigating injury risk. Whilst the presence of intrinsic and extrinsic risks renders the athlete susceptible to injury, these factors are seldom sufficient to produce injury without a mechanism (Bahr and Holme, 2003). Numerous injury aetiology studies (Barker, Beynnon and Renstrom, 1997; Beynnon, Murphy and Alosa, 2002; Murphy, Connolly and Beynnon, 2003, Willems et al., 2005a, b; Engebretsen et al., 2010; Witchalls et al., 2012) have investigated the associated risk factors of ankle pathology. Amongst those identified include, but are not limited to, previous trauma (Tyler et al., 2006; Fulton et al., 2014), neuromuscular patterning (Willems et al., 2002; Fu and Chan, 2005; Willems et al., 2005a, b), functional deficits in strength (Murphy, Connolly and Beynnon, 2003; Willems et al., 2005a, b), limb dominance (Beynnon, Murphy and Alosa, 2002), and fatigue (Gutierrez et al., 2007; Steib et al., 2013; Greig and McNaughton, 2014). The proceeding sections present a review and discussion of the discrete risk factors associated with ankle injury in ballet dancers. Subsequent to a consideration of previous injury, the risk factors thereafter relate to intrinsic, modifiable risk factors.

Previous Injury

There is mounting evidence to suggest that previous injury is one of the strongest predictors of recurrent pathology (Fulton et al., 2014). Although a non-modifiable risk factor, it is important for training and conditioning regimens to account for any enduring functional impairments following initial injury. Whilst many athletes respond well to conservative treatment and often return to competition without issues, residual symptoms have been shown to persist for months and even years, and may lead to long term joint degeneration (Anandacoomarasamy and Barnsley, 2005; Kemler et al., 2016). In a seven-year follow up study following an ankle inversion injury (Konradsen et al., 2002), patients continued to report chronic episodes of pain, swelling and stiffness. The most frequent long-lasting, and perhaps most concerning symptoms following ankle trauma include functional instability and joint laxity, more commonly termed Chronic Ankle Instability (CAI) (van Rijn et al., 2008; Delahunt et al., 2010). The physiological underpinning of CAI stems from deficits in proprioception and strength, and, alterations in muscular recruitment which together induces an overall reduction in joint stability and function, and significantly increases reinjury risk. (Hertel, 2002; van den Bekerom et al., 2013).

Previous research in classical ballet dancers (Leanderson et al., 1996) demonstrated impaired postural stability ‘several’ weeks after the resumption of training and performance, thereby suggesting residual symptoms in proprioception may have been present at the point of return

(Steib et al., 2013). The persistent deficits in sensorimotor control following ankle injury was further evidenced (Hiller, Refshauge and Beard, 2004), with data highlighting greater lateral joint oscillations, and a longer time to stability following a sudden inversion perturbation in the dancers with previous ankle trauma. The high prevalence of compromised ankle function following ankle injury in ballet dancers is reflected in 70% of dancers suffering from CAI (Simon et al., 2014). A possible reason may be that dancers hold back on reporting injury altogether, or temper their description on the severity of symptoms in fear of losing a role/employment (Hamilton et al., 1997). In addition, many dancers don't have access to specialist medical support, and therefore, the rehabilitation process may be inadequate or incomplete (Weiss, Shah and Burchette, 2008). To minimise the risk of recurrent injury, it is vital that adequate and sufficient rehabilitation practices compliment precise injury diagnoses. A critical feature of study design is to control for previous injury through stringent inclusion criteria, which will help to control any confounding results.

External Force and Joint Displacement

Within existing literature, there is a large body of research relating to the association between the forces generated during landing manoeuvres and lower extremity injury (Wang, 2011). The locomotor profile of ballet comprises numerous mechanically demanding jump landing tasks (Twitchett, Koutedakis and Wyon, 2009), with some manoeuvres producing vertical ground reaction forces (GRFs) in excess of 3-4 body weights (Jarvis and Kulig, 2016). Each impact with the ground produces a transient shockwave distally through the body which requires attenuation (Bates et al., 2013). The rate at which the resultant force develops, referred to more commonly as loading rate, is a product of the magnitude of the impact forces and ground impulse, and is implicated in the potential for injury (van der Worp, Vrielink and Bredeweg, 2016). The risk of injury is magnified when acknowledging the limited support and force cushioning properties offered by the ballet shoe (Bickle, Deighan and Theis, 2018). As the primary joint interface between the external forces and the rest of the kinetic chain, the ankle complex is principally responsible for managing the loads imposed by movement (Michael et al., 2008). Ankle joint kinematics therefore have a substantial role in force absorption during landings (Fong et al., 2011). During the stance phase of gait, a position of maximal stability (closed-pack position) corresponds with a dorsiflexed and everted foot alignment (Bonnel et al., 2010). This configuration has also been associated with a reduction in ground reaction forces (DiStefano et al., 2008). However, during jump landing movements, ankle plantarflexion is present at the time of initial contact and is perhaps more profound during ballet-specific manoeuvres given the pointe technique common to performance aesthetics (O'Laughlin, Hodgkins and Kennedy, 2008). The open-pack position of plantarflexion has lower joint congruence and greater instability, which may increase injury susceptibility (Bonnel et al., 2010). In light of these observations, the external forces and excessive ankle joint range of motion inherent to ballet-specific movement present as a plausible risk factor in the high prevalence of ankle injury observed in ballet populations.

An alternate means of quantifying mechanical load in-vivo can be achieved using GPS technology (Boyd, Ball and Aughey, 2013). The integrated inertial component of contemporary devices facilitates a biomechanical perspective of load via calculations from the three-dimensional vector magnitudes of acceleration (Waldron et al., 2011). Segmental loading using accelerometer analysis has been shown to be sensitive to anatomical location of the unit, and greater load responses have been reported between sites, reflecting injury epidemiology observations (Greig and Nagy, 2017; Brogden et al., 2018). Despite the extensive use of GPS technology in performance load assessments, research into the relationship between workload and injury is in its initial stages. Nevertheless, previous research has demonstrated that increases in GPS-derived workloads, specifically excessive acute:chronic workloads over a three-week period, associate with a greater injury risk (Piggott, Newton and Mcguigan, 2009; Colby et al., 2014; Bowen et al., 2017).

Neuromuscular Control

The neuromuscular system governs locomotion and is central to the biomechanics of movement. For the ballet dancer, neuromuscular control is an integral contributor to the processes that underlie postural control and balance (Horak, 2006). During ballet-specific movements, a synchronous activation of the antagonist-agonist muscle actions within the ankle joint is required to maintain correct alignment and stability, and attenuate ground reaction forces (Withrow et al., 2008). Specifically, the peroneal muscles (Longus, Brevis, Tertius) are responsible for providing mediolateral stability and preventing ankle inversion, whilst the Tibialis Anterior and Lateral Gastrocnemius govern plantar- and dorsiflexion function (Bavdek et al., 2018). With the pointe technique common to the multi-planar jump landing tasks of ballet, neuromuscular control is crucial as deficits in ipsilateral and bilateral muscle activation may adversely affect performance and increase injury risk (Batson, 2009).

Existing studies have quantified the neuromuscular contributors to ankle function during discrete techniques such as the plié (Couillandre, Lewton-Brain and Portero, 2008) and relevé (Lin et al., 2016). However, the nature of these floor exercises and the methodological design of the studies suggests that analyses adopted a performance enhancement perspective. Significantly less attention has been directed towards jump landing tasks and the corresponding injury reduction viewpoint. In a study by Lee and colleagues, (2012), the sissonne fermeé technique was used to quantify alterations in muscle activation in ballet dancers with a history of ankle sprain. EMG analyses highlighted that the injured group had a greater co-contraction index compared with uninjured dancers, and therefore, required greater muscle effort to stabilise the ankle joint. The authors inferred that injured dancers adopt a 'load avoidance' strategy to guard against recurrent injury (Lee et al., 2012). The current dearth of information regarding neuromuscular control during ballet-specific jump landing tasks proposes that further research is necessary. Specifically, the plantarflexed and everted/inverted foot configuration characteristic of ballet jump landing tasks, emphasises the importance of investigating the muscular activation strategy used to stabilise the ankle joint.

Joint Strength

In ballet, joint strength in the lower limbs is important towards force production during movement execution, and in facilitating force attenuation during jump landing tasks (Watson et al., 2017). Weakness of the musculature surrounding the ankle complex has long been associated with injury, and the subsequent development of joint instability (Hertel, 2002). Ankle eversion strength, supported by the net peroneal muscles, is fundamental to joint stability and in helping to resist the inversion forces common to injury (Fox et al., 2008). This notion is accentuated given the multi-planar characteristics of ballet-specific movement, and importance of mediolateral stability during landings (Bonnel et al., 2010). Whole dynamic ankle stabilisation requires a coordinated co-activation of all surrounding musculature in all modes of contraction (Kaminski and Hartsell, 2002).

The association between isokinetic ankle strength and injury has been researched previously, but existing findings demonstrate conflicting evidence (Kobayashi and Gamada, 2014). Baumhauer et al., (1995) reported that participants with greater plantarflexion strength but a reduced plantarflexion-dorsiflexion strength ratio sustained more lateral ankle sprains. Further, a similar injury trend was also observed in individuals with a low eversion-inversion strength ratio. Significant eversion torque deficits in both concentric and eccentric modes of contraction have been identified in some studies (Tropp, 1986; Hartsell and Spaulding, 1999; Willems et al., 2002), although the association between eversion-inversion torque and injury hasn't been shown in other research (Porter et al., 2002; Munn et al., 2003, Pontaga, 2004; Sekir et al., 2007). The inconsistent findings, and indeed, lack of available information in ballet populations, suggests that further research on isokinetic ankle strength in dancers is needed. Previous isokinetic methodologies have typically profiled ankle strength at two angular velocities; slow ($30°·s^{-1}$) and fast ($120·s^{-1}$) (Tropp, 1986; Wilkerson, Pinerola and Caturano, 1997; Willems et al., 2002; Pontaga, 2004). However, these testing protocols may be an inadequate approach to quantify muscle strength as a potential risk factor for injury, as sporting locomotion is likely to occur over a range of joint angular displacements and velocities. Hence, isokinetic strength testing protocols ought to be designed and conducted with this in mind, culminating in a functionally relevant methodological approach (Eustace, Page and Greig, 2017).

Limb Dominance

Intra- and inter-limb deficits in strength between contraction modes increases ankle injury risk via compromised support of the joint. In certain sports, athletes may favour the use of a particular leg to execute the technical components of performance, commonly referred to as limb dominance (Daneshjoo et al., 2013). Limb dominance has been implicated as a risk factor for ankle injury due to the potential greater loading on the dominant side (Beynnon, Murphy and Alosa, 2002). The increased frequency and magnitude of moments about the ankle joint may proliferate injury risk, particularly during mechanically demanding activities. However, contradictory findings exist within the limited information available (Murphy, Connolly and Beynnon, 2003). Ekstrand and Gillquist, (1983) conducted one of the earliest investigations in the field and demonstrated that elite male soccer players sustained significantly more ankle injuries in the dominant leg (92%), compared with the non-dominant leg. Conversely, other research found no association between limb dominance and the incidence of ankle pathology (Surve et al., 1994; Beynnon et al., 2001).

Whilst this phenomenon may be more apparent in sports such as soccer (Cuq et al., 2016), it is less obvious in ballet. Yet, dancers may have a preferred leg used for 'pushing off', jumping, and/or landing (Murphy, Connolly and Beynnon, 2003). Bilateral symmetry, particularly in the lower body, is essential in sports with asymmetric kinetic patterns, as prevalent bilateral discrepancies may increase injury risk through compensatory mechanisms involving movement technique and posture (Croisier et al., 2008; Fousekis, Tsepis and Vagenas, 2012; Menzel et al., 2013). In ballet, a dancer is presumed to use each leg equally, and, it is therefore assumed that technique/rehearsal classes prescribe exercises and movements that require equal input from both limbs (Farrar-Baker and Wilmerding, 2006). The importance of bilateral equilibrium is emphasized when considering a typical ballet repertoire contains up to 200 jumps, of which 56% involve one-footed landings (Liederbach et al., 2006).

A study by Mertz and Docherty, (2012) examined bilateral asymmetry in ballet dancers by analysing parameters of postural stability and GRFs when landing from dance-specific jumps. Despite indications of dancers adopting a preferred limb to execute the jump landing tasks, there were no significant bilateral differences in GRFs or balance after landing. Other research in dance has observed the influence of limb dominance from a performance enhancement viewpoint. Lin and colleagues, (2013) assessed the effects of leg dominance on the performance of ballet pirouettes in experienced and novice dancers. The findings revealed that experienced dancers improved performance when using the dominant leg for support, whereas performance was independent of choice of support leg in the novice group. The authors postulated that the significant laterality effect in experienced dancers may be indicative of a chronic adaptation to training dose. Bronner and Ojofeitimi, (2006) however, found no bilateral differences in proficiency when performing the passé, despite an indication of limb preference. In addition, self-reported limb dominance was not reflected in kinematic and kinetic asymmetry during the elevé technique (Abraham et al., 2018). Whilst opposing findings are evident regarding the potential for improved performance when adopting a dominant leg, the presence of limb dominance – potentially causing bilateral asymmetry from the overdevelopment of a particular limb due to asymmetric training – may increase the risk of injury (McHugh et al., 2007).

The conflicting evidence surrounding limb dominance as a risk factor for ankle injury highlights the necessity for further investigation (Ekstrand and Gilquist, 1983; Beynnon et al., 2001). The aforementioned studies represent the few examples of research on limb dominance in ballet yet lack the methodological rigour to propose limb dominance as a potential risk factor for injury. In addition, only one study has adopted an injury reduction perspective via inclusion of a ballet-specific jump landing task. To date, there is limited data available to associate limb dominance as a plausible risk factor for ankle injury in ballet, prompting a need for further studies in a variety of biomechanical factors.

Fatigue

From a mechanical perspective, perhaps the least explored risk factor for injury in ballet is fatigue. The temporal pattern associated with dance injury incidence is well established, yet dancers continue to perform in the absence of adequate rest (Sobrino and Guillen, 2017). Kadel, Teitz and Kronmal, (1992) quantified the injuries sustained by female professional ballet dancers and established that athletes who danced for more than five hours a day, were significantly more likely to sustain injury compared with those who danced less. In addition, other research documented more injuries in the evening, towards the end of the season, and in the latter stages of performance, all proposing fatigue as a contributing factor (Liederbach, Dilgen and Rose, 2008). This association is support by observations in Soccer highlighting that 48% of ankle injuries are sustained in the final third of each respective half (Woods et al., 2003). Considering the musculature surrounding a joint play a pivotal role in dynamic joint stabilisation, it is reasonable to correlate muscular fatigue with an increased risk of injury (Gribble and Hertel, 2004). Enhanced neuromuscular control following fatigue reduces the mechanical stress generated through motion, thereby lessening the amount of force resisted by the articular tissues, and, lowering the risk of trauma (Chappell et al., 2005)

Ballet dancers spend countless hours in technical training in addition to learning, perfecting, and performing choreography, a schedule unconducive to rest (Wyon and Koutedakis, 2013). Due to the nature of dance performance, and the extensive training and competition engagements, fatigue is an unavoidable consequence and places dancers at an increased risk of injury. The activity profile of female professional ballet dancers across a typical workday, highlighted that 90% of the dancers took less than 60 minutes of rest, and ~33% took less than 20 minutes (Twitchett et al., 2010). With ever increasing company pressures to perform in fear of losing a role, dancers may not allow for sufficient recovery, and in many instances train and compete through fatigue and injury (Murgia, 2013). The increased risk of injury resulting from exposure to dance may partially explain the prevalence of injuries attributed to overuse, particularly in the female (70.6%) ballet populations (Smith et al., 2015).

Overuse comprises many factors including poor training and competition scheduling, inadequate physical conditioning, compromised technical execution resulting from neuromuscular deficiencies, and frequent exposure to high impact manoeuvres without sufficient recovery time (Sobrino and Guillen, 2017). In other sports, efforts to reduce the number of injuries attributed to overuse have resulted in governing bodies implementing restrictions on athlete exposure to high-risk locomotion. In cricket, the high incidence of lumbar spine injuries particularly in young fast bowlers has resulted in international legislation advocating limits on training and competitive workloads for bowlers under 21 years of age (Schaefer et al., 2018). However, no such limitations in ballet exist. Over-training, referred to more commonly as 'burnout', is characterised by a sudden drop in performance and often occurs when training and competition 'loads' are matched with inadequate rest (Liederbach et al., 2013). Overtraining is correlated with a higher incidence of injury (Gabbett and Ullah, 2012), hence, an appropriate balance between exposure to dance training and competition workload and recovery, is crucial for maintaining peak performance and minimising injury risk (Gabbett and Domrow, 2007). With a limited understanding on the workload tolerances of ballet dancers, and, that many injuries are attributed to overuse, further research in ballet is required. Accelerometers offer an attractive method in quantifying the mechanical demands of ballet, which may inform the strategies aimed at monitoring dancers' physical and technical development, whilst potentially identifying markers for injury.

Summary

The mechanism of ankle injury is well described, with jump landing manoeuvres consistently identified as the prime locomotor contributor to injury. The intrinsic risk factors for ankle injury are multi-faceted and less well understood, especially in ballet given the lack of research in these populations. Investigations into the discrete risk factors mentioned above requires the use of a range of biomechanical tools of analysis. The subsequent section will identify and critically discuss the use of the relevant technologies in existing ballet research.

2.3 Biomechanical Analyses in Ballet

This section presents an appraisal of the use of biomechanics equipment within existing investigations in ballet. Studies that have incorporated automated motion capture and/or force plate analysis, EMG, isokinetic dynamometry or accelerometry are included and discussed under the relevant sub-headings. For the automated motion capture and force plate analysis, and EMG sections, only articles published since December 2009 are referred to as to prevent duplicating the information presented in a previous review on biomechanics research in ballet (Krasnow et al., 2011). Isokinetic dynamometry and accelerometry were absent from the specified review, and thus, all pertinent investigations are included. The proceeding text will summarise and emphasise trends within current ballet biomechanics research, and discuss the limitations of methodological design which will inform the development of the experimental chapters comprising the current thesis.

Automated Motion Capture and Force Plate Analysis

Kinematic and kinetic analyses are a vital component of sports performance monitoring, providing information on the displacement, velocity and acceleration of body segments, and, the magnitude of external force acting on the body during motion (Wilson and Kwon, 2008; Hood, McBain and Portas, 2012). Kinematic and kinetic analyses in ballet are a burgeoning line of enquiry, with advancements in motion capture resulting in multi-camera systems and the capacity to synchronously obtain ground reaction force parameters (DiCesare et al., 2014). This technology facilitates sophisticated biomechanical assessments, enabling researchers to quantify the segmental contributors to discrete movements towards performance excellence, whilst identifying locomotor patterns associated with injury risk (Krasnow et al., 2011). Table 2.6 presents a summary of the studies in ballet which have conducted kinematic and/or kinetic evaluations. Corresponding information relates to the participant cohort, and the methodological design with reference to the task of analysis used and the kinematic and kinetic variables obtained. Whilst varying factors are considered across the specified investigations, analyses are encompassed under 5 discrete movement categories; gait and standing posture, ballet floor exercises, drop landing tasks, jump landing tasks, and ballet-specific jump landing tasks. The following text will allude to the specific research foci of relevant studies and critically discuss the methodological designs.

Table 2.6. Articles comprising a kinetic and/or kinematic analysis of ballet movement with key methodological information presented.

Authors, Year	**Participants, n, Gender (if specified), Skill-level (if specified) Age.**	**Task of Analysis**	**Biomechanical Measures**
Orishimo et al., 2009	33 – 12 male; 21 female, professional 26 ± 4.5 years	Single leg drop landings (30 cm)	Peak vertical GRFs, Loading Rate; Sagittal and frontal plane hip knee and ankle joint kinematics/moments
Bruyneel et al., 2010	40 – 12 males 28 females, professional, 17.5 ± 3.51 years	Standing posture	3D GRFs
Imura et al., 2010	7, female, experienced, 27.7 ± 1.7 years	Fouetté turn	3D hip, knee and ankle kinematics
Hackney et al., 2011	13, female, university-level, 21.31 ± 2.06 years	Grand jeté	Vertical GRFs; Sagittal plane hip, knee and ankle displacement (joint stiffness)
Kulig et al., 2011	12, 6 males; 6 females, pre-professional, 18.9 ± years	Saut de chat	Vertical GRFs; Sagittal plane knee kinematics
Lin et al., 2011	33 – 22 dancers (11 injured, 11 uninjured); 11 non-dancers, females, 19.5 ± 2.47 years	Single leg standing (1st and 5th position, en pointe)	Anteroposterior and mediolateral GRF oscillations (COP); 3D hip, knee and ankle kinematics
Pappas et al., 2011	36 – 13 males, 23 females, professional, age not specified	Vertical jump	Anteroposterior and mediolateral GRFs (impulse)
Walter et al., 2011	18, female, college-level, 19.94 ± 1.16	Assemblé	Vertical GRFs
Bronner, 2012	27, 11 males; 16 females, pre-professional, 21.43 ± 0.66 years	Developpé arabesque	Pelvic and gesture limb displacement
Fietzer et al., 2012	18 – 12 healthy, 6 with patellar tendinopathy, 50 % gender split, pre-professional, 18.88 ± 0.96 years	Saut de chat	Anteroposterior GRFs; Sagittal plane hip, knee and ankle discplacment.

Table 2.6. *Continued*

Kilby and Newell, 2012	20 – 10 dancers; 10 regular exercising non-dancers, female, experienced, 21.25 ± 2.45 years	Single and double leg quiet stance	Anteroposterior and mediolateral GRF oscillations (COP)
Krasnow et al., 2012	40, female, mixed-ability, 30.0 ± 13.0 years	Grand battement	Pelvic and trunk centre of gravity displacement
Lee et al., 2012	22 – 11 injured; 11 uninjured, university-level, 19.24 ± 2.77 years	Sissonne fermeé	Frontal plane ankle and foot kinematics
Mertz and Docherty, 2012	33 – 7 males; 23 females, university- level, 19.6 ± 1.1 years	Changement, Entrechat trois	Vertical GRFs, Anteroposterior and mediolateral GRF oscillations (COP)
Lin et al., 2013	26, female, mixed ability, 14.89 ± 2.65 years	Pirouette	Vertical GRFs; Sagittal plane hip, knee and ankle kinematics.
Volkerding et al., 2013	15 – 8 dancers; 7 non-dancers, college-level, 20.7 ± 0.79 years	Single leg drop landing	Vertical GRFs; Sagittal plane hip, knee and ankle kinematics
Kim et al., 2014	9 – 1 male; 8 females, college-level, 23.0 ± 2.9 years	Pirouette	Upper body angular momentum and range of motion
Lin et al., 2014	26, female, mixed ability, 14.89 ± 2.65 years	Pirouette	Sagittal and Transverse plane hip, knee and ankle kinematics
Lin et al., 2014	33 – 13 injured; 20 uninjured, female, college-level; 18.35 ± 2.30 years	Grand plie	Anteroposterior and mediolateral GRF oscillations (COP); Pelvic motion, sagittal plane knee and ankle kinematics
Liederbach et al., 2014	80 – 40 dancers; 40 team sport athlete, 50% gender split, college-level, 23.5 ± 3.75 years	Single- leg drop landing (30 cm)	Sagittal and frontal plane hip and knee joint moments; Sagittal and frontal plane trunk, hp and knee kinematics

Table 2.6. *Continued*

Orishimo et al., 2014	80 – 40 dancers; 40 team sport athlete, 50% gender split, college-level, 23.5 ± 3.75 years	Single leg drop landing (30 cm)	Sagittal and frontal plane hip and knee joint moments; Sagittal and frontal plane trunk, hp and knee kinematics
Tepla et al., 2014	25 – 13 dancers (5 males; 8 females); 12 controls (3 males; 9 females), professional, 24.15 ± 3.28 years	Walking	3D pelvis, hip, knee and ankle kinematics.
Gontijo et al., 2015	20, experienced, 27 ± 8 years	Plie	Pelvic and midfoot stability, alignment of knee and ipsilateral foot
Hendry et al., 2015	18, female, mixed ability, 12-15 years	Vertical jumps with turnout; unilateral, first and second position	Peak posterior knee shear forces; Sagittal plane hip and knee kinematics
Hopper et al., 2015	13 – 5 males; 8 females, experienced-professional, 20.7 ± 5.1 years	Single leg drop landing (20 cm)	Vertical GRFs; Sagittal plane ankle kinematics
Peng et al., 2015	25 – 11 injured; 14 – uninjured, female, university-level, 18.25 ± 0.45 years	Échappé	Vertical GRFs; Sagittal plane hip, knee and ankle joint stiffness
Casabona et al., 2016	20 – 10 dancers; 10 untrained, females, professional, 25.55 ±	Standing (five foot positions)	Anteroposterior and mediolateral GRF oscillations (COP)
Jarvis and Kulig, 2016	30, experienced, 20.8 ± 4.9 years	Saut de chat	Vertical GRFs, impulse; Sagittal plane hip, knee, ankle and metatarsophalangeal kinematics
Jarvis and Kulig, 2016	10, female, professional, 18-35 years	Relevés, Sautés, Saut de chat	Sagittal plane metatarsophalangeal joint moments; Sagittal plane metatarsophalangeal joint kinematics
Zaferiou et al., 2016	11, female, skilled, 21.7 ± 3.72 years	Pirouette	3D hip, knee and ankle net joint moments

Table 2.6. *Continued*

Imura and Lino, 2017	12, female, professional, 30.0 ± 1.0 years	Countermovement jump	Sagittal and frontal plane hip, knee and ankle joint moments
de Mello Viero et al., 2017	17, female, experienced, 12-35 years	Demi-plié, Grand battement, (grand) Jeté	Midfoot longitudinal height, hip external rotation
Quanbeck et al., 2017	10, female, mixed ability, 20.3 ± 1.5 years	Turnout	Transverse plane hip motion, femoral anteversion, transmalleolar angle
Tanabe, Fujii and Kouzaki, 2017	7, female, non-professional, 24.10 ± 5.00 years	Tiptoe standing	Sagittal plane hip, knee, ankle and metatarsophalangeal kinematics
Zaferiou et al., 2017	5, female, skilled, age not specified	Pirouette	3D hip, knee and ankle net joint moments; Angular momentum kinematics
Bickle et al., 2018	15, female, professional, 26 ± 4 years	Bourres	Vertical GRFs; Sagittal plane ankle and midfoot kinematics
Bruyneel, et al., 2018	40 – 12 males; 28 females, professional, 17.5 ± 3.51 years	Grand plié	3D GRFs, COP
Carter et al., 2018	6, female, university-level, 18.8 ± 0.8 years	Turnout plie, turnout risk, flex-point-flex	3D multi-segmental foot kinematics
Carter et al., 2018	22, university-level, 19.1 ± 1.8 years	Turnout	Transverse hip kinematic, 3D foot kinematics
Imura and Lino, 2018	8 – 6 male; 2 females, professional, 15-42 years	Pirouette	Upper and lower body angular momentum

Table 2.6. *Continued*

Harwood et al., 2018	30 – 13 dancers; 17 non-dancers, female, amateur, 11.35 ± 0.85 years	Single leg vertical hop, stop-jump task	Vertical GRFs; 3D hip, knee and ankle joint moments/kinematics
Michalska et al., 2018	26 – 13 dancers; 13 controls, experienced, 25.5 ± 5 years	Quiet standing	Anteroposterior and mediolateral GRFs (COP)
Turner et al., 2018	27 – 12 dancers; 15 non-dancers, female, experienced, 18-25 years	Double leg drop landing (30.5 cm)	Fontal plane hip and knee kinematics
Azavedo et al., 2019	30 – 15dancers; 15 non-dancers, experienced, 25.8 ± 6 years	Single leg lateral., diagonal, forward jumps	Vertical GRFs; Sagittal and frontal plane hip and knee joint moments/kinematics
Carter et al., 2019	18, female, university-level, 18.8 ± 1.6 years	Turnout, Sautés	3D multi-segmental foot kinematics
Swain et al., 2019	60 – 21 dancers; 39 non-dancers, female, 15+ years	Standing and sitting	Frontal plane lumbar flexion in standing, transverse plane in sitting
Ward et al., 2019	79 – 39 dancers , elite; 40 temp-sport athletes, college-level, 50% gender split, 23.92 ± 4.40 years	Single leg drop landing (30 cm)	Sagittal plane hip, knee and ankle joint moments

n, number; GRFs. Ground Reaction Forces; COP, Centre of Pressure.

In gait and posture focused studies, motion capture and/or force plates have predominantly been used to quantify postural stability via examination of GRF oscillations (centre of pressure – COP) and joint displacements during general (Bruyneel et al., 2010; Tepla et al., 2014; Michalska et al., 2018; Swain et al., 2019) and ballet-specific (Lin et al., 2011; Casabona et al., 2016; Bickle, Deighan and Theis, 2018) tasks. The focus for some research has been to investigate stability differences between younger and older athletes (Bruyneel et al., 2010), and between ballet dancers and non-dance populations (Tepla et al 2014; Michalska et al., 2018; Swain et al., 2019), whilst others have investigated the effects of foot position (Casabona et al., 2016) and shoe age (Kilby and Newell, 2010; Bickle, Deighan and Theis, 2018). The aesthetic component towards ballet performance success has understandably led to extensive research into the kinematic and kinetic characteristics of discrete ballet floor techniques. Selected tasks incorporated into relevant studies include the grand plié (Lin et al 2014; Gontijo et al., 2015; Bruyneel, Bertrand and Mesure, 2018), fouetté turn (Imura, Lino and Kojima, 2010), the pirouette (Lin et al 2013; Kim et al., 2014; Lin et al., 2014; Zaferiou, Wilcox and McNitt-Gray, 2016; Zaferiou et al., 2017; Imura and Lino, 2018) and the developpé arabesque (Bronner, 2012). These studies have distinguished and differentiated the kinematic and kinetic responses between limbs (Imura, Lino and Kojima, 2010), movements with and without turnout (Zaferiou, Wilcox and McNitt-Gray, 2016, 2016; Quanbeck et al., 2017; Zaferiou et al., 2017; Carter et al., 2018; Carter, Bryant and Hopper, 2019), and task execution between dancers of varying skill level (Bronner, 2012; Krasnow et al., 2012; Lin et al., 2013; Lin et al., 2014). Analyses of gait during general tasks, and ballet floor exercises, adds value to the understanding of the segmental contributions towards movement. However, the studies mentioned above predominantly adopt a performance enhancement focus, and the specified tasks negate an understanding of the kinematic and kinetic implications of landings towards an increased injury risk (Walter, Docherty and Schrader, 2011).

Biomechanical assessments of drop landing tasks in ballet dancers have increased in recent times, with drop boxes or platforms set at a 30 cm height comprising most investigations. Drop landing analyses have been conducted to investigate dance floor properties on ankle kinematics (Hopper et al., 2015),whilst within-ballet gender comparisons (Orishimo et al., 2009) and comparisons between sport modalities (Orishimo et al., 2014; Liederbach et al., 2014; Ward et al., 2019) have been the focus for other studies. Orishimo et al., (2009) revealed no significant differences in sagittal and frontal plane joint kinematics at the hip, knee and ankle between female and male professional ballet dancers during a unilateral drop landing task. In addition, gender failed to significantly influence peak vertical GRFs, vertical loading rate and net joint moments (Orishimo et al., 2009). There is growing research interest surrounding variations in injury epidemiology between difference sports. Specifically, the considerably lower prevalence of Anterior Cruciate Ligament (ACL) pathology in ballet dancers compared with athletes competing in other sports. There is conflicting evidence regarding a disparity in peak vertical GRFs during drop landing tasks between sport modalities (Volkerding and Ketcham, 2013; Ward et al., 2019). However, relevant studies corroborate on kinematic observations, with dancers demonstrating significantly greater hip and knee flexion angles, and significantly decreased knee valgus alignment compared with their non-dancing counterparts (Orishimo et al., 2014; Turner et al., 2018). These observations suggest that dancers adopt safer landing strategies in terms of the common ACL injury mechanism (near-to-full knee extension coupled with valgus), which may partially explain the variance in ACL incidences between populations. Whilst comparator studies provide valuable information on injury aetiology to non-dance populations, they neglect a focus on the injuries specific to ballet. In addition, although drop landing tasks serve as an appropriate compromise in studies comparing athletes from difference sports, the pre-defined height selection and anticipated execution of the task is unrepresentative of typical ballet movement. Hence, methodologies comprising a drop landing task may hinder the interpretation on the kinematic and kinetic demands of ballet towards understanding injury risk.

In contrast to drop landing tasks, jump landing manoeuvres account for the aerial component preceding the ground contact phase, but have received comparatively less attention despite their closer association with the ballet movement profile. The vertical jump was used to compared time to stability differences upon landing on flat and raked floors (Pappas et al., 2011), and, to investigate the effects of turnout on hip and knee joint torques (Imura and Lino, 2017). Kinetic responses and lower extremity kinematics have also been quantified during a maximal vertical hop, and, a stop jump task (Harwood et al., 2018). The findings showed no significant difference in vertical GRFs between dancers and non-dancers, but dancers had significantly greater sagittal plane hip, knee and ankle joint excursions (Harwood et al., 2018). More recently, Azavedo and colleagues, (2019) examined the influence of multi-directional landings on lower limb biomechanics. Analyses showed that jump direction had a significant effect on resultant joint alignment, specifically, that the lateral jump task elicited significantly greater hip abduction moments and ankle eversion, and, lower hip flexion and ankle plantarflexion compared with the forward and diagonal variants. The evident influence of jump direction on lower limb biomechanics supports the inclusion of multi-planar tasks within biomechanical assessments of ballet to differentiate the mechanical implications.

To maximise ecological validity in biomechanics research, the tasks incorporated into investigations should simulate those observed during sporting performance. For ballet, these are the discrete jump landing tasks allied to a choreographed routine. Within existing literature, jumps, hops and leaps feature in numerous kinematic and kinetic studies to examine the effects of factors including flooring (Hackney et al., 2011), limb dominance (Mertz and Docherty, 2012), footwear (Walter, Docherty and Schrader, 2011; McPherson, Schrader and Docherty, 2019), taping conditions (Hendry et al., 2015), and previous injury (Fietzer, Chang and Kulig, 2012; Lee et al., 2012; Peng et al., 2015). Whilst the échappé (Peng et al., 2015) and sissonne (Le et al., 2012) have formed some methodological designs, the saut de chat leap is one of most commonly used in biomechanical assessments of ballet-specific jump landings (Kulig, Fietzer and Popovich, 2011; Fietzer, Chang and Kulig, 2012; Jarvis and Kulig, 2016). Kulig and colleagues, (2011) examined kinetic response and knee mechanics during the take-off and landing phase of a saut de chat leap. During the weight acceptance phase, peak vertical GRFs were significantly greater in the landing component (4.38 ± 0.82 N·kg^{-1}) compared with the take-off phase (3.48 ± 0.36 N·kg^{-1}), as was vertical mean loading rate (62.57 vs 34.8 BW/s). Kinematic analyses revealed a 67% increase in knee flexion during the landing phase potentially owing to the greater GRFs present. Proceeding research supports the kinetic observation of greater GRFs during the landing phase of the saut de chat (Jarvis and Kulig, 2016). However, the kinematic findings highlighted that the take-off phase places greater kinematic demand on more distally located joints, whereas the proximal joints exhibit greater kinematic demand during the landing component. Repetition of these mechanically demanding jump landing tasks during an extensive training period may contribute to the high frequency of chronic injuries sustained at the ankle in ballet populations.

Despite evident growth in the quantity of kinematic and kinetic studies conducted, and the availability of data in ballet dancers, issues persist within the experimental designs of contemporary investigations. Numerous methodologies comprise a uni- or bilateral drop landing task, or, clinically focused vertical jumps (Orishimo et al., 2009,Orishimo et al., 2014; Liederbach et al., 2014; Turner et al., 2018; Ward et al., 2019). Though conducted from an injury aetiology perspective, such tasks have limited association with ballet movement. Furthermore, the between-sport comparison design may highlight important clinical information for the populations with greater injury incidences, but this approach ignores the epidemiology of injuries specific to ballet. Vertical jumps and countermovement jumps have improved specificity towards ballet performance (Pappas et al., 2011; Imura and Lino, 2017; Harwood et al., 2018), but they negate the anteroposterior and mediolateral aerial displacement, and, the multi-planar landing characteristics inherent to many ballet techniques. Consequently, resultant kinetic responses and lower limb joint configurations are likely very different from those of the discrete ballet manoeuvres. The jump landing tasks of a typical routine fit the narrative of the five jump descriptors, are multi-planar in nature, and contain varying levels of jump height, all factors which influence resultant joint kinematics and kinetics. Some researchers have incorporated ballet-specific manoeuvres into assessments of lower limb mechanics which more closely simulate true ballet movement, and improves ecological validity (Kulig, Fietzer and Popovich, 2011; Fietzer, Chang and Kulig, 2012, Lee et al., 2012; Peng et al., 2015, Jarvis and Kulig 2016). However, these tasks are typically analysed in isolation, predominantly considered unilaterally, and cover only a small proportion of the techniques contained within a ballet performance repertoire. In addition, the land and hold instruction is quite different to the landing strategies observed during a routine, whereby dancers are often required to land and rapidly transition into a connecting movement of a sequence. Existing biomechanics studies in ballet therefore refute the complex multi-planar movement profile of ballet, and thus, there is limited practical application beyond the specific methodological design used.

Electromyography

Electromyography (EMG) records the electrical activity elicited by a muscle in response to a contraction (Vigotsky et al., 2018). It is a popular analysis tool in biomechanics research, enabling evaluations on the functional status of skeletal muscles. In a review of biomechanics research in ballet (Krasnow et al., 2011), EMG has been used extensively to quantify the muscle actions contributing to techniques including the plié (Ferland, Gerinder and Lébe-Néron, 1983; Trepman et al., 1994, 1998; Couillandre, Lewton-Brain and Portero, 2008), relevé (Massó et al., 2004; de Bartolomeo et al., 2007), degagé (Mouchnino et al., 1992; Lepelley et al., 2006), developpé (Wilmerding et al., 2001), and the grand battement (Ryman and Ranney, 1978). The predominant theme that emerged from the EMG-inclusive studies of the Krasnow and colleagues, (2011) review was that comparisons of neuromuscular activation between elite dancers, novice dancers, and non-dancers are commonplace. The information demonstrated that dancers, generally, have superior motor strategies compared with those of their non-dancing counterparts, and, elite dancers have more proficient neuromuscular patterning compared with non-elite dancers (Krasnow et al., 2011). Table 2.7 relates to each study that has incorporated EMG into analyses. Presented within, is information pertaining to the methodological approach and the EMG metrics quantified during the research. There is a vast range of tasks used within the relevant articles, and the level of EMG analysis varies from a 2-electrode, to a 16-electrode design. Four studies have evaluated bilateral muscle responses, three unilaterally in the dominant limb and one in the non-dominant limb.

Table 2.7. A methodological summary of relevant EMG-inclusive studies in ballet biomechanics research.

Author(s) (Year)	Participants, n, Gender (if specified), Skill-level (if specified), Age.	Task of Analysis	Electrode Placement	Outcome Measure
Krasnow et al., 2012	40, mixed- ability ballet or modern, 30 ± 13 years	Grand battement	Bilateral: RA, AH, ES, GA, GM, BF, Quads (unspecified), TA	% MVIC
Lee et al., 2012	22 – 11 injured; 11 uninjured, university-level, 19.24 ± 2.77 years	Sissonne fermeé	Bilateral: PL, TA, MG, VL, HA, Hams	RMS normalised to Median magnitude of trial, Co-contraction Index.
Lin et al., 2014	33 – 13 injured; 20 uninjured, female, college-level; 18.35 ± 2.3 years	Grand plie	Bilateral: PL, MG, TA	Mean EMG
Lin et al., 2016	20, female, college-level, 17.98 ± 1.51 years	Relevé en demi-pointe	Non-dominant: TA, PL, MS, MG, Hams, RF	% MVC
Tanabe, Fujii and Kouzaki, 2017	7, female, non-professional, 24.1 ± 5 years	Tiptoe standing	Dominant: GM, RF, SR, VL, BF, SM, MG, LG, SL, PL, TA, ED, FH	Mean EMG, EMG coherence (coactivation)
Zaferiou et al., 2017	5, female, skilled, age not specified	Pirouette	BF, GM, GME	% MVIC
Saito et al., 2018	20 – 11 trained; 9 untrained, female, 20.7 ± 1.6 years	Relevé to max plantarflexion	Dominant: SL, MG	Mean EMG
Turner et al., 2018	27 – 12 dancers; 15 active non-dancers, female, experienced, 18-25 years	Double leg drop landing (30.5 cm)	Bilateral: GM, GME	% MVIC
Aquino et al., 2019	9, female, professional, 22.2 ± 2.2 years	Relevé and arabesque	Dominant: TA, MG	% MVC

n, number; RA, Rectus Abdominus; AH, Abductor Hallicus; ES, Erector Spinae; GA, Gastrocnemius; GM, Gluteus Maximus; GME, Gluteus Medius; BF; Bicep Femoris; Quads, Quadriceps; TA, Tibialis Anterior; PL, Peroneus Longus; MG, Medial Gastrocnemius; VL, Vastus Lateralis; Hams, Hamstrings; SL, Soleus; RF, Rectus Femoris; SR, Sartorius; SM, Semimembranosus; ED, Extensor Digitorum; FH, Flexor Hallicus; MVIC, Maximal Voluntary Isometric Contraction, MVC, Maximal Voluntary Contraction.

Several research themes are evident in the studies conducting EMG analyses. Some research has examined joint coordination and muscular contributions to tiptoe standing (Tanabe, Fujii and Kouzaki, 2017), whilst others have quantified EMG responses towards executing a ballet-specific technique (Zaferiou et al., 2017). A between-skill level comparison focus was adopted during an EMG evaluation of the trunk and lower limb muscles during the grand battement devant (Krasnow et al., 2012), whereas Saito and colleagues, (2018) revealed the unique and superior neuromuscular strategy of ballet dancers compared with non-dancers during the relevé. The nature of these studies suggests a performance enhancement rationale owing to the description of the muscular contributors to simple and functionally relevant tasks, and, the inclusion of ballet dancers of varying ability or a control group. This aesthetic focus negates an understanding on the injury risk implications from neuromuscular control.

From an injury management standpoint, the relevé has been used to investigate the effects of fatigue (Lin et al., 2016) and shoe wear (Aquino et al., 2019) on resultant lower limb EMG response, and the grand plié, to examine the influence of previous injury (Lin et al., 2014). The pirouette, relevé, and grand plie represent ballet floor exercises, in that there is no elevation component. The high incidence of ankle injuries in ballet has been associated with repetition of the high impact jump landing tasks comprising a typical routine. EMG analysis of floor exercises may limit the understating on injury risk given the greater kinetic demands of jump landing manoeuvres. Turner et al., (2018) used a drop landing task, but this movement has limited specificity towards ballet, and EMG analysis was conducted on the gluteal muscles in accordance with ACL injury risk. To date, only Lee and co-authors, (2012) have incorporated a ballet-specific jump landing task. The sissonne fermeé was used to demonstrate that previously injured dancers have altered Peroneus Longus, Tibialis Anterior and Gastrocnemius muscle activation compared with uninjured dancers. Identifying deficits in neuromuscular control post injury may be crucial in preventing reoccurrence. However, there is limited consideration of EMG responses towards initial injury, not least, during a battery of tasks designed to reflect the high kinetic and multi-planar demands of ballet movement. Acknowledgement of this, and an EMG focus on the musculature governing ankle function is required in further research.

Isokinetic Dynamometry

Isokinetic dynamometry is the gold standard method of assessing joint torque, providing insight into muscle strength capacity during motion (Munoz-Bermejo at al, 2019). Isokinetic dynamometers enable joint strength to be evaluated for each muscle contraction mode, and across a range of angular velocities and joint displacements (Evangelidis, Pain and Folland, 2015). There is a wide application of isokinetic strength assessment in sport, with the data informing training interventions and the rehabilitation process towards return to play (Undheim et al., 2015). For a ballet dancer, strength has a pivotal role in force generation and management during the execution of discrete routine elements (Watson et al., 2017). Nevertheless, Table 2.8 highlights that isokinetic strength data in ballet populations is relatively sparse compared with other sporting modalities. Preliminary observations of Table 2.8 show an overriding focus on the knee joint for strength assessments. However, there is variety in the methodological approach with reference to the contraction modes evaluated , and the angular velocities used. Of the strength metrics available for analysis, peak torque is the primary outcome measure used in relevant ballet studies.

Table 2.8. Relevant articles comprising an isokinetic strength assessment of ballet dancers.

Authors (Year)	Participants, n, Gender (if specified), Skill-level (if specified) Age.	Isokinetic protocol, Reps/Sets, Angular velocity (if specified)	Outcome measure
Westblad et al., 1995	52 – 17 dancers (6 males; 11 females); 35 active controls (17 males; 18 females), professional, 30.25 ± 5.75 years	ECC and CON knee extensors, single-action test – 3 maximal reps; repetitive actions test – 100 reps , $90°·s^{-1}$	Peak torque, ECC:CON total work ratio.
Koutedakis, 1997	42 – 20 males; 22 females, professional, 26.85 ± 5.7 years	CON knee flexion and extension, 3 reps/1 set, $60°·s^{-1}$ and $240°·s^{-1}$	Peak torque
Koutedakis and Sharp, 2004	22, female, professional, 25 ± 1.3 years	CON knee flexion and extension, 3 reps, $60°·s^{-1}$	Peak torque
Thomas and Parcell, 2004	45 – 15 ballet; 15 folk; 15 non-dancers, university level, 21 ± 0.3 years	ISO and CON Plantarflexion, 1 rep/1set for ISO and CON at each angular velocity; 30 reps for CON fatigue, ISO (80, 90, $100°·s^{-1}$), CON (60, 90, 120, 180, 240 and $300°·s^{-1}$), CON fatigue ($300°·s^{-1}$)	Peak torque, Fatigue index
Gupta et al., 2004	71 – 34 dancers; 37 non-dancers, female, professional, 19.45 ± 1.5 years	CON hip external rotation, not specified, $30°·s^{-1}$	Peak torque, Angle-specific torque
Kenne and Unnithan, 2008	21 – 11 dancers, 10 basketballers, females, experienced, 15.8 ± 1.1 years	CON and ECC knee flexion and extension; plantar- and dorsiflexion, not specified, 30, 60 and $90°·s^{-1}$	Peak torque
Lim et al., 2015	45 – 32 dancers; 13 controls, female, university level, 18-20 years	CON knee flexion and extension, 3 reps/1 set, $60°·s^{-1}$	Peak torque
Tsanaka et al., 2017	17, female, college level, 22.88 ± 3.10 years	CON knee flexion and extension, 3 reps at $60°·s^{-1}$; 5 reps at $180°·s^{-1}$	Peak torque

Table 2.8 *Continued.*

Lima et al., 2018	27 – 12 dancers; 15 resistance trained, female, experienced, 23.35 ± 2.42 years	CON knee flexion and extension, 5 reps/1 set, 60°·s^{-1}	Peak torque, H:Q ratio

n, number; reps, repetitions; ECC, Eccentric; CON, Concentric; ISO, Isometric °·s^{-1}, degrees per second; H:Q, Hamstrings to Quadriceps.

Isokinetic strength assessments of the musculature governing knee joint function dominate the relevant studies (Westblad, Tsai-Fellander and Johansson, 1995; Koutedakis, 1997; Koutedakis and Sharp, 2004; Kenne and Unnithan, 2008; Lim et al., 2015; Tsanaka, Manou and Kellis, 2017; Lima et al., 2018), with limited data available for hip (Gupta et al., 2004) and ankle joint strength (Thomas and Parcell, 2004; Kenne and Unnithan, 2008) respectively. The research foci of the relevant isokinetic strength studies include associations between knee joint strength and lower limb injury (Koutedakis, 1997), comparisons between dance and non-dance populations (Westblad, Tsai-Fellander and Johansson, 1995; Kenne and Unnithan, 2008), and dancers training in various genres (Lim et al., 2015). There is also an evaluation on the effects of stretching modalities (Lima et al., 2018), and targeted strength training interventions (Koutedakis and Sharp 2004; Tsanaka, Manou and Kellis, 2017) on knee flexor and extensor torque.

The strong emphasis on the knee joint of the aforementioned studies may indeed reflect the importance of knee flexor and extensor strength towards ballet performance. Muscle contraction type and angular velocity are important considerations when developing an isokinetic testing protocol (Eustace, Page and Greig, 2017). The concentric evaluation comprising many investigations may have been selected in consideration of the many jumping manoeuvres observed in a ballet routine, to which high knee flexor and extensor torque generation is crucial. However, significantly less attention has been direction towards eccentric strength despite is role in force attenuation during landing (Jones et al., 2017). Hence, the majority of current isokinetic strength testing methodologies align with the performance enhancement strand of biomechanics research. The angular velocities of the relevant studies appear to have been selected without functional consideration, and typically represent a slow or fast speed. For example, only two studies have included a testing velocity exceeding $90°\cdot s^{-1}$ (Koutedakis, 1997; Tsanaka, Manou and Kellis 2017). Knee extensor and flexor angular velocities during a jump landing manoeuvre are likely to far exceed the slower velocities observed in most studies, and thus, current methodologies may not adequately reflect the functional characteristics in strength required during ballet movement.

Another key component of isokinetic strength analyses are the resultant metrics obtained. In the relevant ballet studies, peak torque is the most commonly reported measure, providing a single maximum torque value from a pre-determined range of motion at a specified angular velocity (Morel et al., 2015). Peak torque values can be used to quantify antagonistic muscle pair strength ratios to identify potential issues with ipsilateral joint strength and function, which are presented in two studies (Westblad, Tsai-Fellander and Johansson 1995; Lima et al., 2018). Whilst peak torque is valuable for quantifying absolute maximal joint strength, it provides no information on the angular displacement at which it is achieved, nor, how strength is maintained over the range of motion. Angle-specific torque is presented in one study (Gupta et al., 2004), which represents the only example to have considered isokinetic strength beyond the traditional peak torque metric. With the angle at which peak torque is achieved having been shown to differ between contraction modes (Small et al., 2010), consideration of angle-specific derivatives of strength and the implications to dynamic strength ratios is advocated (El-Ashker et al., 2017).

There is an evident shortage in ankle joint strength assessments in ballet dancers, despite epidemiology research identifying the ankle as the prime injury location (Smith et al., 2015). Further, the ankle is responsible for absorbing and managing the forces allied to ground contact (Michael et al., 2008), to which multi-planar strength has a vital role in reducing injury risk (Fox et al., 2008). Current isokinetic methodologies are restricted to measures of plantar- and dorsiflexor strength perhaps given the number of relevés and (demi)pliés performed during a routine. However, the eversion and inversion alignment common to the aetiology of ankle injuries have yet to be included in isokinetic strength testing protocols. To that end, a comprehensive assessment of ankle eversion and inversion strength in ballet is required to inform the strategies implemented in training, and, the knowledge surrounding injury occurrence. However, this may only be achieved by quantifying strength in multiple modes of contraction, over several joint angular displacements, and using a range of angular velocities (Evangelidis, Pain and Folland, 2015; Eustace, Page and Greig, 2017).

Accelerometry

Perhaps the least utilised biomechanical tool of analysis in ballet research is accelerometry. Whilst there is an abundance of literature surrounding the application of accelerometry to movement analyses particularly in team invasion sports such as Australian Rules Football (Cormack et al., 2013), Soccer (Scott et al., 2013) and Rugby (Gabbett, Jenkins and Abernethy, 2010; Suarez-Arrones et al., 2012), there is a lack of information available in a dance context. Potential predictors of mechanical loading and performance in dancers were investigated in a series of experimental studies (Armstrong et al., 2018; Armstrong et al., 2019; Armstrong, Brogden and Greig, 2020). The findings demonstrated that the deep squat, hurdle step, and the functional movement screen composite score, can be used to predict mechanical loading during dance performance (Armstrong et al., 2018), but the star excursion balance test (Armstrong et al., 2019) and joint hypermobility (Armstrong, Brogden and Greig, 2020) are not good predictors of mechanical loading during dance. Accelerometer-derived load responses were investigated in dancers during a performance-specific movement protocol comprising multiple stages of progressive intensity (Brogden et al., 2018). The study found that increases in load responses were commensurate with increases in exercise intensity and duration. Further, accelerometer load was significantly greater in the lower limb compared with responses measured using the conventional C7 unit location. These observations add value in demonstrating the sensitivity of accelerometry to dance-specific movement, and in highlighting the influence of unit placement on resultant load which has important clinical implications when interpreting performance demands. With this study representing the first insight into the workload tolerances of ballet dancers, current understanding of performer capacity towards the rigours of ballet is limited.

In recent times, several research projects have attempted to quantify associations between accelerometer-derived metrics and GRFs, with conflicting findings reported (Wundersitz et al., 2013; Nedergaard et al., 2017: Edwards et al., 2019). Only one article has adopted this approach in ballet performance (Almonroeder et al., 2019), which examined the relationship between pelvic-worn accelerometer impact accelerations and peak vertical GRFs and corresponding loading rates. Further, the study investigated whether accelerometers are sensitive to fatigue-induced changes in landing forces. In the study, ballet dancers were required to land on a force platform whilst performing consecutive changement de pied until self-determined exhaustion (defined as the point at which they could no longer perform the jump with correct technique). The means for peak vertical accelerations, peak vertical GRFs and peak instantaneous loading rates were calculated for the first 10 landings (baseline), and thereafter at the landing corresponding with 25%, 50% and 75% of the total jump count until the final 10 landings (100% time point). Correlation analyses revealed very strong, positive relationships between impact accelerations and peak vertical GRFs ($r \geq 0.95$, $p < 0.01$), and between impact accelerations and instantaneous loading rates ($r \geq 0.80$, $p < 0.01$) for all landing time points (Almonroeder et al., 2019). These findings highlight that wearable accelerometry could be used to estimate landing forces and loading rates in ballet, and that this method of analysis provides a promising performance monitoring tool when quantifying landing impacts during ballet performance. The innovative methodological approach offers scope for subsequent research in ballet, and although in its infancy, very strong, positive relationships between impact accelerations and corresponding GRFs offer a promising alternative to classic laboratory-based assessments of loading. The tri-axial nature of accelerometry mirrors the traditional 3D vector magnitudes of force plate analysis, and thus, suits the complexity of ballet movement. Furthermore, accelerometers can be worn during ballet choreography thereby increasing the application of mechanical analysis beyond laboratory research. Accelerometer-based microtechnologies might prove a valuable assessment tool in the screening and monitoring of ballet dancers towards injury reduction and/or during rehabilitation.

2.4 Summary

Current methodological designs in biomechanical investigations of ballet movement may be hindering the understanding of injury occurrence. Research ought to be designed in accordance with the epidemiology specific to ballet (i.e. the high prevalence of ankle injuries). Despite an increase in the number of studies incorporating ballet-specific techniques, relevant kinematic and kinetic analyses have typically used a solitary manoeuvre to examine the mechanical implications of movement. Future investigations should look to design a battery of ballet-specific tasks, reflecting the multi-planar characteristics of the ballet movement repertoire, and with a consideration of uni- and bilateral landings that differentiate between holding and transitioning. The level of biomechanical analysis may also be improved via synchronous consideration of EMG, thereby quantifying the neuromuscular strategies governing movement during these tasks. The lack of available information on isokinetic ankle strength, particularly in eversion and inversion, warrants further attention. Subsequent methodologies ought to consider the inter-relationships between contraction mode, angular velocity and joint displacement, and, additional strength metrics to peak torque. Finally, current ballet performance monitoring practices may be strengthened through accelerometry. Recent evidence has demonstrated positive relationships between accelerometry and ground reaction force measures in jump-landing tasks. The limited research on this relationship during a range of ballet-specific movements suggests that this research strand requires further exploration, and subsequent findings may offer scope to implementing biomechanical evaluations in field-based designs.

Chapter 3. General Methodology

3.1 Identification and Eligibility of Participants

To conform with the aims and objectives of the current thesis, all participants involved in each study were amateur female ballet dancers. Participant sample size and information relating to demographics is contained within each experimental chapter. Amateur status was defined as participants not training or competing as part of a professional organisation. A random stratified sampling method was deployed to recruit dancers from existing Edge Hill University undergraduate populations, and external groups utilising the same facilities. Both participant suitability and eligibility to take part in the research was determined against strict inclusion criteria. Prospective participants conformed to the inclusion criteria if they were aged 18+, had a minimum of 8 years dancing experience, and were currently active in ballet (attending ballet training for a minimum of three hours per week). Exclusion criteria included any neurological, visual or vestibular disorder, or a history of severe cardiovascular and/or pulmonary disease. Moreover, participants were prohibited from taking part if they had sustained a lower limb musculoskeletal injury in the six months before the testing date, or deemed to have chronic ankle instability as determined by the Cumberland Ankle Instability Tool (CAIT) questionnaire (Appendix 1). The CAIT is a 9-item scale filled out for both ankles, enabling an explicit assessment of each limb. The questions contain a multiple answer option, and the CAIT is able to quantify the severity of ankle instability with a numerical value. Resultant scores are out of 30, and a cut-off of 27.5 has been recommended to discriminate between stable and chronically unstable ankles (Hiller et al., 2006). The CAIT is a valid and reliable (ICC = 0.96) tool to measure the severity of functional ankle instability.

Ethical Considerations

In accordance with Edge Hill University's policy on undertaking research with human subjects, each study obeyed the ethical application process and was granted approval by the Department of Sport & Physical Activity's ethics committee prior to any data collection. Approval of appropriately completed risk assessments allied to each experimental practice was also obtained. Involvement in the research was entirely voluntary, and all participants provided written informed consent in adherence with the principles outlined in the

Declaration of Helsinki. To ensure consent was fully informed, all participants received a detailed information sheet. Included were the precise study protocols, the requirements of the experimental trial from the participant, and any risks associated with completion of the exercise. Agreement to take part in the research reserved each participants' right to withdraw at any stage during their testing session, and up to four weeks after data collection. All equipment required for each experiment was risk assessed for hazards that may increase risk, and, calibrated according to the manufacturer's specifications.

Pre-Data Collection Measures

All participants provided their age (years), whilst height (cm) and mass (kg) were logged using a wall-mounted stadiometer (Holtain, Harpenden, HSK-BI, UK) and top pan scales (Seca, Germany) respectively. Further eligibility for inclusion in the research was quantified using a series of pre-exercise screening assessments. All participants completed a comprehensive medical questionnaire to provide information relating to previous illnesses, history of musculoskeletal injury, and any health disorders that may prohibit their suitability to take part (Appendix 2). Both resting heart rate and blood pressure were measured (Omron, Mx3 plus, Netherlands) with values of < 90 bpm and 140/90 mmhg required for subsequent participation. Any participant whose values exceeded these limits, or indeed presented with any medical issue, were unable to complete the testing. Signs of physical and/or mental distress during the data collection process were monitored throughout.

3.2 Experimental Procedures

All experimental trials were conducted between 10:00-17:00 in accordance with the typical daily schedule of a ballet dancer (Twitchett et al., 2010). Data collection was conducted in a laboratory and/or dance studio with ambient temperature and humidity approximating 21.5° and 35 g/m^3 respectively. Each participant was advised to attend their testing session in a euhydrated and post-absorptive state. Furthermore, participation in each study required a 24-hour abstinence from vigorous exercise and alcohol consumption. Attire similar to that worn during a typical ballet training session was recommended, however personal ballet shoes were a necessity. Prior to each data collection session and for each study, all participants completed a standardized warm-up protocol targeting ankle joint mobilisation. Ballet-specific exercises (10 x plié, 10 x relevé (heel raise), 5 x arabesque for each limb) were followed by a blend of self-desired dynamic and static stretches. Where appropriate, additional warm-up components relating to the specific exercise mode were completed, and are described in detail within the relevant experimental chapters.

Ballet-specific Jump Landing Battery

A ballet-specific jump landing testing battery comprising seven tasks was designed in accordance with the discrete techniques of a typical repertoire, and, the tri-planar nature of biomechanics. Each movement was carefully selected to fit within the narrative of the five basic jump descriptors (one foot to the same foot; one foot to the opposite foot; one foot to two feet; two feet to one foot; two feet to two feet). The manoeuvres used in the testing protocol were the jeté, jeté step, échappé, sissonne, sissonne pas de bourres (PDB), temps levé and jeté en tournant (ET), and Table 3.1 provides a supplementary technique descriptor for each task.

Table 3.1. A glossary of terms with corresponding technique descriptors for each movement of the ballet-specific movement battery.

Jump landing Manoeuvre	Execution Description
Jeté (One foot to the opposite foot)	A leap characterised by a shift in body weight from one foot to the other. The dancer displaces the legs in a front ‘split’ formation and holds upon landing.
Jeté step (One foot to the opposite foot)	The dancer executes the jeté using the technique described above. However, upon landing, the dancer ‘steps out’ thereby simulating the typical landing component of a sequence.
Échappé (Two feet to Two feet)	A two-part movement starting in second position and jumping out into second position. The dancer is instructed to hold the second position upon landing.
Sissonne (Two feet to one foot)	Initiated in fifth position, the dancer executes a lateral jump from two feet to one and holds upon landing.
Sissonne pas de bourres (Two feet to one foot)	The dancer completes the sissonne technique outlined above, however transitions into Pas de Bourres immediately on landing.

Table 3.1. *Continued.*

Temps levé (One foot to the same foot)	A small hop on one foot, with the contralateral foot raised off the floor.
Jeté en Tournant (One foot to the opposite foot).	The dancer completes a jeté technique with an additional 180° horizontal rotation during the aerial aspect of the jump. The landing position is held.

Each manoeuvre was selected to distinguish mechanical variations between tasks, with a consideration of linear vs rotational demand, and, single- vs double-leg landings. The jeté, échappé, sissonne, temps levé and jeté ET required a land and hold terminal position to replicate the clinical tasks such as single- and double-leg drop landings observed in biomechanical analyses (Pozzi et al., 2017). However, the jeté step and sissonne PDB contained a transitional landing component to simulate the movement profile typical of ballet following a jump landing manoeuvre. The testing battery was therefore designed to mediate a balance between ecological validity and experimental control. Figure 3.1 presents a montage of the jump-landing manoeuvres demonstrating each stage of execution, as described in Table 3.1.

Figure 3.1. A montage representing the technique required for each jump landing task.

Selected tasks from the above movement battery are embedded within a choreographed routine in a later experimental chapter (study 6) with further detail contained therein. Thus, there is a progression from analysing discrete tasks in a laboratory setting, to evaluating load responses to a performance piece.

3.3 Methodological Approach to the Problem

The following section contextualises the approach for the current thesis, underlining the rationale for the methodological constructs of each experimental chapter. Central to the methodological approach is the importance of analysing movements specific to ballet, and the cohesion and synergy of a multi-modal research design to reflect the breadth and capacity of biomechanical analysis tools. A further consideration of improved ecological validity was outlined by acknowledging methods that facilitate field-based assessments of biomechanics. The epidemiology of injuries in ballet highlights a high prevalence of overuse type traumas particularly in the amateur female population (Smith et al., 2015), with injuries primarily localised to the foot and ankle complex (Gamboa et al., 2008; Liederbach, Dilgen and Rose, 2008; Bowerman et al., 2014). The principle mechanism of injury was determined as jump landing manoeuvres (Fousekis, Tsepis and Vagenas, 2012), but injury risk has a multi-factorial profile (Murphy, Connolly and Beynnon, 2003, Willems et al., 2005 a, b; Engebretsen et al., 2010; Witchalls et al., 2012) and therefore warrant a multi-factorial investigation. Hence, there is vast opportunity to develop ballet research using a variety of biomechanical techniques, as evident in the experimental chapters of the current thesis.

The battery of ballet-specific jump landing tasks described in Section 3.2 are investigated using laboratory-based motion capture and force plate analysis during study 1. In accordance with injury epidemiology (Smith et al., 2015) and aetiology (Wang, 2011), 3D ankle kinematics and vertical kinetic parameters are quantified. Between-task comparisons are conducted, providing a biomechanical profile of ballet-specific movement. Bilateral responses are also considered in accordance with the asymmetric movement patterns typical of ballet. Study 2 evaluates the muscle activity of the Peroneus Longus, Tibialis Anterior and Lateral Gastrocnemius to the same battery of ballet movements. Bilateral EMG responses to each task are discussed in relation to the kinetic and kinematic findings of study 1, initiating a synthesis of biomechanical analysis tools which develops through the thesis. Study 3 presents a rigorous assessment of ankle eversion/inversion strength using isokinetic dynamometry, again with a consideration of laterality. The specific testing protocol used was supported by the coronal plane angular velocity, and sagittal and coronal plane angular displacement data obtained during study 1. Isokinetic strength metrics used for analysis include peak torque, and more contemporary variables such as functional range, angle-specific torque, and dynamic control ratios, providing a more in-depth evaluation of strength capacity in ballet dancers. Bilateral EMG responses of the same muscles used in study 2 data are also collected during the isokinetic testing protocol, enabling a function vs capacity consideration of EMG.

Study 4 introduces a portable method of investigating biomechanics during the same movement battery via use of tri-axial accelerometry. Total load (defined as PlayerLoad^{TM}), and the relative planar contributors to load for each movement are quantified, with accelerometry differentiating the kinematic characteristics inherent to each task. In each experimental chapter, the findings are interpreted in relation to the preceding methodologies, building a layering and multi-factorial appreciation of the biomechanical demands of this movement battery. The synthesis of biomechanical analysis tools is highlighted in study 5, which explores an association between accelerometry and vertical GRFs, with a focus on the utility and efficacy of accelerometry for valid assessments of biomechanics in field-based environments. To prevent ethical issues relating to over-testing individual participants, data collection for kinematics/kinetics (study 1), EMG (study 2), and accelerometry (study 3) occurred simultaneously during a single testing session. Isokinetic strength testing (study 4) was also conducted in the same visit to the laboratory, but followed a one-hour rest period to prevent fatigue compromising performance and resultant data.

The final experimental study quantifies mechanical load responses to an authentic, ballet-specific choreographed routine using tri-axial accelerometry. Inclusion of additional accelerometers to supplement the traditional C7 placement, and to provide greater anatomical relevance to the preceding focus on lower limb injury, increases rigour in the assessment of ballet performance with implications for current athlete monitoring practices. Figure 3.1 provides an illustration of the biomechanical research process undertaken by the current thesis. The laboratory investigations are designed with cohesion and demonstrable synergy between the various analysis tools used in biomechanics. The structure of the discrete experimental studies was developed in consideration of a transition between laboratory- and field-based research, but with a logical and evident rationale.

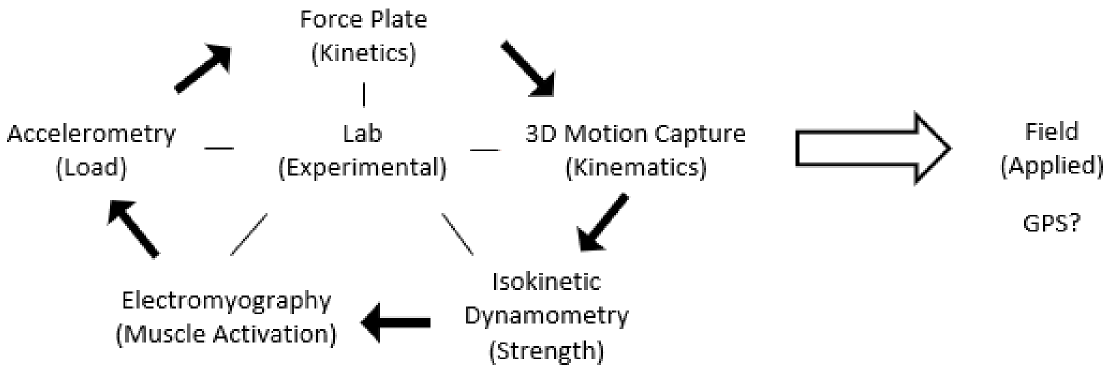

Figure 3.2. A schematic of the thesis research process.

Chapter 4. Study one: The Kinematic and Kinetic Responses to Ballet-specific Movement.

4.1 Introduction

A classical ballet performance repertoire comprises numerous mechanically demanding jump landing manoeuvres (Twitchett, Koutedakis and Wyon, 2009). Whilst success is largely dependent on grace and aesthetic merit (Hincapie, Morton and Cassidy, 2008), the complexity of the ballet movement profile carries an inherent susceptibility to injury (Smith et al., 2015). Injury epidemiology surveillances in ballet have established lower extremity injury rates ranging between 59%-93% (Bowerman et al., 2014; Ramkumar et al., 2016), with up to 57% of these injuries affecting the foot and ankle complex (Smith et al., 2015). Jump landing tasks or movements involving rapid deceleration and/or a change in direction are regarded as the primary locomotor contributors to the ankle injury mechanism (Fousekis, Tsepis and Vagenas, 2012). The pointe technique employed in ballet is a pivotal criterion of success, but increases the mechanical complexity of jump landing techniques and exacerbates the plantarflexion alignment common to the injury mechanism (O'Laughlin, Hodgkins and Kennedy, 2008). Ballet-specific manoeuvres are characterised by high ground reaction forces (GRFs) and multi-planar joint displacements, factors which pose a risk of injury to performers (Wang, 2011).

Existing biomechanical analyses in ballet have typically used clinical tasks such as single- and double-leg drop landings from pre-determined heights (Orishimo et al., 2009; Volkerding and Ketcham, 2013; Liederbach et al., 2014; Turner et al., 2018), however, these have limited biomechanical validity in relation to the explosive jump landing elements characteristic of ballet performance (Walter, Docherty and Schrader, 2011). Where ballet-specific techniques have been incorporated into relevant studies, biomechanical analyses have been applied to a solitary movement (Kulig, Fietzer and Popovich, 2011; Lee et al., 2012; Peng et al., 2015; Jarvis and Kulig, 2016). Although valuable, and with evident improvement in ecological validity, a single manoeuvre design fails to consider the varying technical components involved in ballet performance. In addition, the land and hold instruction common to many investigations is markedly different to the ballet movement profile, whereby a landing is frequently followed by a rapid transition into another technique (Twitchett, Koutedakis and Wyon, 2009).

The current study therefore presents a novel battery of ballet-specific jump landing manoeuvres designed to highlight the technical characteristics associated with linear and rotational movements. The testing protocol also differentiates between landings performed uni- or bilaterally, and between landings characterised as 'land and hold' to arrest momentum, and 'land into transition' to simulate moving into a connecting sequence. The aim of the study was to quantify kinetic and ankle joint kinematics responses – due to greater injury incidence observed at this joint (Smith et al., 2015) – to a ballet-specific movement battery. The aesthetic requirements of ballet performance are reflected in training prescription that emphasises equal exposure to physical and technical demands from both limbs (Farrar-Baker and Wilmerding, 2006). Furthermore, aetiological information relating to limb dominance/preference and implications on injury susceptibility is limited and currently equivocal. Hence, an additional aim was to investigate bilateral symmetry in the biomechanical response to each ballet-specific movement.

Experimental research questions:

1) Does the identification of a dominant limb reflect in the biomechanical response to ballet-specific jump-landing tasks?
2) Does task-specificity (i.e. 'land and hold' vs 'land into transition' influence the kinetic and kinematic responses to ballet-specific movements?

4.2 Methods

Participants

An *a priori* power calculation was conducted from pilot study data using G*Power software (v 3.1, Heinrich-Heine-Universität, Dusseldorf, Germany), with measures in peak vertical GRF. The power analysis revealed that a sample size of 14 was required to elicit an observed statistical power of 0.8, and to determine significant main effects and interactions between limbs for the dependent variables. To mitigate ethical issues surrounding over testing, a sample of 14 female ballet dancers (Age: 19.29 ± 1.59 yrs; Height: 1.65 ± 0.05 m; Body mass: 61 ± 8.29 kg) were recruited. All participants actively volunteered to participate and provided written informed consent. Eligibility to participate was subject to each participant conforming with the inclusion and exclusion criteria outlined in the general methodology section (Chapter 3.1). Pre-test health screening and anthropometric measures were also completed according to protocol.

Experimental Procedures

All participants were required to attend the Biomechanics Laboratory at Edge Hill University on one occasion to complete the experimental testing procedures. Participants wore apparel consistent with ballet training and/or performance, and their own ballet shoes. Prior to data collection, participants completed the standardised warm up protocol described in Chapter 3. Within the warm-up procedure, participants also completed two familiarisations trials of the battery of ballet-specific movements outlined in Chapter 3 in the order contained therein, for both the dominant and non-dominant limb. Limb dominance was determined by asking each participant to identify which limb they prefer to land on during a unilateral ballet-specific jump landing task, in accordance with previous methods (Mertz and Docherty, 2012; Carcia, Cacolice and McGeary, 2019). The familiarisation trials helped to establish a start position, and in mitigating targeting of the force platform during the experimental trials. Subsequent to the warm-up process, all participants were instructed to perform three repetitions of each ballet-specific movement for the dominant and non-dominant leg, adopting the same technique as used during ballet performance Six trials of each movement were completed before progressing on to the next to enable the participant to develop their technical strategy for each manoeuvre. During the experimental trials, the order of tasks and starting limb was randomised between participants.

All participants initiated each respective trial as to achieve a full foot contact on to a single embedded force platform (Model 9281CA, Kistler National Instruments, Winterthur, Switzerland; Dimensions: length ≈ 60 cm, width ≈ 40 cm), configured to sample data at 2000 Hz. The échappé, sissonne, sissonne PDB and temps levé started one stride length from the force plate. The jeté, jeté step and jeté ET contained an approach phase, however, during the respective trials, all participants were required to start these movements within 60 cm of each other to limit the influence on resultant data. A ten-camera motion capture system (Oqus 300, Qualisys Medical AB, Gothenburg, Sweden) set to sample data at 250 Hz, was employed to capture the position of retroreflective markers placed on the body using passive infrared technology. Prior to each individual data collection session, the laboratory global co-ordinate system and data capture volume area were delineated. The origin of the global co-ordinate system was determined using a static reference L-frame containing four marker locations. The desired capture volume was achieved by waving a T-shaped wand of known length (749.9 mm) through the anticipated motion area. The norms of residuals and the standard deviation of the known wand length following a 45-second dynamic calibration of the motion capture system, were inspected to check for any errors with either the camera system, or the spatial positioning of the markers. Calibrations were deemed acceptable when residuals of < 0.5mm and points above 3000 in all cameras were produced. Kinematic and kinetic data were recorded synchronously, and interfaced using Qualisys Track Manager (QTM) software.

Quantifying 3D Kinematics

Lower limb 3D segmental modelling and tracking were achieved using the Calibrated Anatomical System Technique (CAST) marker configuration (Cappozzo et al., 1995). The CAST technique enables segments to be modelled in 6 degrees of freedom by identifying an anatomical frame for each segment from selected anatomical locations. The modelling of a single rigid foot, and shank segment, was achieved using retroreflective markers (8mm) taped bilaterally on to the ballet shoe, and were attached superficial to the 1st, 2nd and 5th metatarsal heads, calcaneus, medial and lateral malleoli, and the medial and lateral femoral epicondyles. The ankle joint co-ordinate system axes were determined as the mid-point between the medial and lateral malleoli markers, whilst the knee joint co-ordinate system axes were defined as the mid-point between the positions of the medial and lateral epicondyle markers. The foot segment was tracked using the 1st, 2nd and 5th metatarsal heads, and the calcaneus markers, and the shank via a carbon fibre casing comprising four retroreflective markers strapped to the lateral aspect of the mid-shank. The tracking markers/cluster configurations during the jump landing (dynamic) trials were referenced to an initial standing static calibration.

Data Processing

Post-processing was conducted in QTM, initiating with all jump landing trials being cropped to the window of interest as to discard any irrelevant motion data. Anatomical marker identification was completed using a combination of manual, and the Automatic Identification of Markers (AIM) model methods. Any marker temporarily occluded within the capture volume area resulted in data gaps and reduced the accuracy of tracking. Instances of marker occlusion resulted in the use of in-built gap-filling methods to reconstruct the position of the marker within the lab, and maximise data quality. Processed QTM files were subsequently converted to C3D files and exported to Visual 3-D software (C-Motion Inc., Germantown, Maryland, USA) for further processing.

Marker displacement data was initially smoothed using a zero-lag, 4th order, low-pass Butterworth filter with a 12 Hz cut-off frequency enabling ankle joint kinematics to be calculated. GRF data was similarly smoothed using a cut-off frequency of 100 Hz in accordance with previously established methods (Sugiyama et al., 2014). Initial contact (IC) for each jump landing manoeuvre was determined as the first point > 20 N of applied vertical force on the platform. For the jump-hold techniques, the stance phase was calculated as the period between IC and body weight stabilisation, defined as the time taken for the vertical GRF signal to stabilise within 5% of the participants' body weight following landing from the jump (Flanagan, Ebben and Jenson, 2008). The stance phase of the transition movements was determined as the time between IC and Toe-off (TO), identified by the first instance of < 20 N of applied vertical force. To minimise planar crosstalk, an XYZ cardan sequence of rotation was adopted in the calculation of angular kinematics (Sinclair et al., 2012). Stance phase kinematic measures selected for statistical analysis were: 1) Angle at footstrike; 2) Peak angle; 3) Relative range of motion (expressing the angular displacement between angle at footstrike and peak angle. For the current investigation, 0° represents full plantarflexion in the sagittal plane, with values over 90° indicative of dorsiflexion. In the coronal (inversion/eversion), and transverse (internal-/external rotation) planes, 0° signifies a neutral foot alignment. Kinetic parameters were; 1) Impact vertical peak force ($_{V}$GRF); 2) Time to impact vertical peak force; 3) Mean vertical loading rate ($_{V}$LR, calculated as impact vertical peak force/time to impact vertical peak force) (Crowell and Davis, 2011). GRF and vLR data were normalised to participant body weight (BW) to enable between-limb comparisons across participants, and the movement battery. Test-retest reliability for each of the biomechanical variables was measured using intra-class correlations coefficients (ICCs) with 95% confidence intervals (CI). Findings from reliability analysis for kinetic and kinematic data are contained within Tables 4.1 and 4.2 respectively, with reliability thresholds (Poor < 0.40; Fair = 0.40-0.70; Good =0.70-0.90; Excellent = > 0.90) interpreted according to Coppieters et al., (2002).

Table 4.1. Measures of reliability (ICC – (95% CI)) for kinetic data across the movement battery.

		Jump-Landing Manoeuvre						
Variable	**Limb**	**Jeté**	**Jeté Step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
Peak vGRF ($N \cdot kg^{-1}$)	Dominant	0.89 (0.70-0.97)	0.96 (0.91-0.99)	0.95 (0.87-0.98)	0.93 (0.84-0.98)	0.96 (0.90-0.99)	0.97 (0.93-0.99)	0.93 (0.84-0.98)
	Non-Dominant	0.88 (0.70-0.96)	0.94 (0.84-0.98)	0.89 (0.74-0.96)	0.95 (0.88-0.98)	0.97 (0.92-0.99)	0.96 (0.90-0.99)	0.95 (0.88-0.98)
Time to peak vGRF (s)	Dominant	0.83 (0.59-0.94)	0.92 (0.82-0.97)	0.92 (0.80-0.97)	0.86 (0.67-0.95)	0.82 (0.56-0.94)	0.91 (0.79-0.97)	0.95 (0.87-0.98)
	Non-Dominant	0.82 (0.55-0.94)	0.76 (0.27-0.92)	0.71 (0.28-0.90)	0.90 (0.76-0.97)	0.73 (0.31-0.91)	0.92 (0.80-0.97)	0.91 (0.78-0.97)
Mean vLR ($BW \cdot s^{-1}$)	Dominant	0.89 (0.67-0.96)	0.95 (0.88-0.98)	0.95 (0.87-0.98)	0.91 (0.77-0.97)	0.93 (0.82-0.97)	0.90 (0.75-0.96)	0.88 (0.72-0.96)
	Non-Dominant	0.86 (0.66-0.95)	0.91 (0.78-0.97)	0.80 (0.51-0.93)	0.88 (0.71-0.96)	0.96 (0.91-0.99)	0.90 (0.76-0.97)	0.90 (0.75-0.97)

$N \cdot kg^{-1}$, Newtons per kilogram; s, Seconds; $BW \cdot s^{-1}$, Body weights per second; PDB, Pas de Bourres; ET, en Tournant.

Table 4.2. Measures of reliability (ICC – (95% CI)) for kinematic data across the movement battery.

Sagittal		**Jump-Landing Manoeuvre**						
Variable	**Limb**	**Jeté**	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
Angle at FS	Dominant	0.92 (0.80-0.97)	0.96 (0.90-0.99)	0.90 (0.76-0.97)	0.94 (0.86-0.98)	0.90 (0.77-0.97)	0.94 (0.86-0.98)	0.98 (0.94-0.99)
(°)	Non-Dominant	0.98 (0.96-0.99)	0.96 (0.91-0.99)	0.96 (0.89-0.98)	0.93 (0.82-0.97)	0.92 (0.82-0.97)	0.98 (0.94-0.99)	0.97 (0.92-0.99)
Peak Angle	Dominant	0.92 (0.81-0.97)	0.90 (0.75-0.96)	0.88 (0.72-0.96)	0.93 (0.84-0.98)	0.97 (0.93-0.99)	0.87 (0.67-0.95)	0.94 (0.86-0.98)
(°)	Non-Dominant	0.90 (0.74-0.96)	0.92 (0.78-0.97)	0.93 (0.82-0.97)	0.90 (0.75-0.96)	0.94 (0.86-0.98)	0.96 (0.90-0.99)	0.91 (0.78-0.97)
RROM	Dominant	0.93 (0.84-0.98)	0.94 (0.86-0.98)	0.71 (0.33-0.90)	0.94 (0.86-0.98)	0.96 (0.91-0.99)	0.91 (0.78-0.97)	0.92 (0.82-0.97)
(°)	Non-Dominant	0.87 (0.69-0.96)	0.82 (0.56-0.94)	0.84 (0.60-0.94)	0.81 (0.55-0.93)	0.92 (0.80-0.97)	0.95 (0.89-0.98)	0.86 (0.67-0.95)
Coronal								
Angle at FS	Dominant	0.83 (0.58-0.94)	0.94 (0.85-0.98)	0.93 (0.84-0.98)	0.94 (0.86-0.98)	0.95 (0.88-0.98)	0.84 (0.61-0.95)	0.86 (0.67-0.95)
(°)	Non-Dominant	0.93 (0.83-0.98)	0.95 (0.87-0.98)	0.85 (0.62-0.95)	0.94 (0.85-0.98)	0.80 (0.37-0.94)	0.91 (0.79-0.97)	0.97 (0.92-0.99)
Peak Angle	Dominant	0.73 (0.34-0.91)	0.91 (0.77-0.97)	0.85 (0.63-0.95)	0.93 (0.84-0.98)	0.80 (0.52-0.93)	0.94 (0.85-0.98)	0.81 (0.53-0.93)
(°)	Non-Dominant	0.95 (0.87-0.98)	0.94 (0.85-0.98)	0.87 (0.67-0.95)	0.92 (0.81-0.97)	0.95 (0.88-0.98)	0.94 (0.85-0.98)	0.91 (0.77-0.97)
RROM	Dominant	0.61 (0.07-0.86)	0.90 (0.51-0.97)	0.64 (0.07-0.88)	0.84 (0.63-0.95)	0.72 (0.34-0.90)	0.90 (0.75-0.97)	0.79 (0.52-0.94)
(°)	Non-Dominant	0.92 (0.79-0.97)	0.84 (0.60-0.94)	0.89 (0.75-0.96)	0.81 (0.55-0.93)	0.85 (0.64-0.95)	0.88 (0.72-0.96)	0.78 (0.47-0.92)
Transverse								
Angle at FS	Dominant	0.97 (0.91-0.99)	0.96 (0.89-0.99)	0.96 (0.91-0.99)	0.97 (0.93-0.99)	0.95 (0.88-0.98)	0.99 (0.96-1.00)	0.93 (0.82-0.97)
(°)	Non-Dominant	0.96 (0.88-0.98)	0.90 (0.76-0.97)	0.91 (0.79-0.97)	0.91 (0.78-0.97)	0.88 (0.71-0.96)	0.91 (0.78-0.97)	0.92 (0.82-0.97)
Peak Angle	Dominant	0.89 (0.72-0.96)	0.91 (0.79-0.97)	0.94 (0.86-0.98)	0.96 (0.90-0.99)	0.87 (0.67-0.95)	0.94 (0.86-0.98)	0.96 (0.89-0.99)
(°)	Non-Dominant	0.94 (0.94-0.86)	0.97 (0.93-0.99)	0.93 (0.83-0.98)	0.86 (0.67-0.95)	0.86 (0.65-0.95)	0.90 (0.75-0.96)	0.95 (0.88-0.98)
RROM	Dominant	0.72 (0.29-0.90)	0.86 (0.65-0.95)	0.76 (0.40-0.92)	0.61 (0.18-0.87)	0.67 (0.18-0.89)	0.58 (0.03-0.85)	0.66 (0.17-0.88)
(°)	Non-Dominant	0.70 (0.29-0.90)	0.91 (0.77-0.97)	0.81 (0.52-0.93)	0.63 (0.14-0.87)	0.79 (0.58-0.93)	0.75 (0.36-0.91)	0.82 (0.58-0.94)

FS, footstrike; °, degrees; RROM, relative range of motion; PDB, Pas de Bourres; ET, en Tournant.

Statistical Analysis

Data was analysed using IBM SPSS Statistics V25.0 software (IBM, Armonk, New York, USA). Descriptive statistics (mean ± σ) are presented for each kinematic and kinetic variable across the seven-movement testing battery. With the data normality assumption confirmed via histograms and a Shapiro-Wilk test, a 7 x 2 (movement x limb) repeated measures Analysis of Variance (ANOVA) was employed to investigate main effects for movement and laterality, and the movement *x* laterality interaction for each of the specified outcome measures. Post-hoc pairwise comparisons with a Bonferroni correction factor were conducted for any identified significant main effects and interactions, and 95% Confidence intervals (CI) and Cohen's *d* effect sizes (small, 0.20-0.49; moderate, 0.50-0.79; large, ≥ 0.80) were presented (Cohen, 1988). Differences were deemed statistically significant at the $p < 0.05$ level.

4.3 Results

Kinetic Parameters

Table 4.3 summarises the influence of movement task and laterality on the kinetic response. Significant between-movement differences were identified for peak $_V$GRF ($F = 13.40$, $p < 0.01$), time to peak $_V$GRF ($F = 22.94$, $p < 0.01$), and mean $_V$MLR ($F = 7.76$, $p = 0.02$). For these dependent variables, the values reported in the subsequent text represent an average from the dominant and non-dominant limb, and this is consistent throughout. There was a significant main effect for limb observed in peak vertical force ($F = 4.85$, $p = 0.04$), with greater values revealed in the dominant limb, however this was not highlighted in time to peak $_V$GRF ($F = 0.01$, $p = 0.94$) or mean $_V$LR ($F = 2.64$, $p = 0.13$).

Table 4.3. Bilateral comparison of selected Kinetic variables during the seven dance-specific movement battery. Corresponding values are mean ± σ.

Variable	Limb	Jump-Landing Manoeuvre						
		Jeté	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
Peak vGRF ($N \cdot kg^{-1}$)*	Dominant	$2.99 \pm 0.79^{\delta 3}$	$3.04 \pm 1.06^{\delta 3}$	1.92 ± 0.69	$2.67 \pm 0.80^{\delta 3}$	$2.44 \pm 0.66^{\delta 3}$	$2.91 \pm 0.79^{\delta 35}$	$3.23 \pm 0.82^{\delta 35}$
	Non-Dominant	$2.87 \pm 0.65^{\delta 3}$	$2.67 \pm 0.79^{\delta 3}$	1.75 ± 0.56	$2.58 \pm 0.64^{\delta 3}$	$2.48 \pm 0.77^{\delta 3}$	$2.98 \pm 0.70^{\delta 35}$	$2.95 \pm 0.82^{\delta 35}$
Time to peak vGRF (s)	Dominant	$0.07 \pm 0.02^{\delta 56}$	$0.06 \pm 0.02^{\delta 56}$	$0.07 \pm 0.01^{\delta 56}$	$0.10 \pm 0.02^{\delta 1237}$	$0.09 \pm 0.01^{\delta 1235 \Phi}$	$0.10 \pm 0.01^{\delta 1236 \Phi}$	$0.08 \pm 0.02^{\delta 12}$
	Non-Dominant	$0.07 \pm 0.01^{\delta 56}$	$0.07 \pm 0.01^{\delta 56}$	$0.07 \pm 0.01^{\delta 56}$	$0.10 \pm 0.01^{\delta 1237}$	$0.08 \pm 0.01^{\delta 1235 \Phi}$	$0.10 \pm 0.01^{\delta 1236 \Phi}$	$0.09 \pm 0.02^{\delta 12}$
Mean vLR ($BW \cdot s^{-1}$)	Dominant	50.83 ± 27.51	55.59 ± 36.55	$30.68 \pm 15.55^{\delta 37}$	31.47 ± 15.72	29.38 ± 13.11	32.29 ± 15.15	$44.90 \pm 19.72^{\delta 3}$
	Non-Dominant	46.63 ± 17.59	44.33 ± 20.45	$26.19 \pm 11.01^{\delta 37}$	30.21 ± 13.30	30.32 ± 14.23	33.47 ± 14.94	$36.15 \pm 14.50^{\delta 3}$

$N \cdot kg^{-1}$, Newtons per kilogram; s, Seconds; $BW \cdot s^{-1}$, Body weights per second; PDB, Pas de Bourres; ET, en Tournant. * denotes a significant difference between limbs irrespective of movement. Φ denotes a significant difference between limbs for a movement. δ denotes a significant main effect for movement irrespective of limb, and ^{1234567} signify which movements are significantly different from each other (^{1}) is jeté and (^{7}) is jeté ET.

Peak Vertical Force

Post-hoc analyses following the significant main effect for movement revealed that peak $_{V}$GRF was significantly lower during the échappé (1.84 ± 0.63 $N \cdot kg^{-1}$; CI: 1.50-2.17) compared with all other movements ($p \leq 0.01$, d = 0.42-0.65). In addition, peak $_{V}$GRF during the jeté ET (3.09 ± 0.82 $N \cdot kg^{-1}$; CI: 2.63-3.54) and temps levé (2.94 ± 0.75 $N \cdot kg^{-1}$; CI: 2.48-3.40), was significantly greater than the sissonne PDB (2.46 ± 0.72 $N \cdot kg^{-1}$; CI: 2.05-2.87, $p \leq 0.03$, d = 0.31-0.48). Peak $_{V}$GRF was significantly greater in the dominant limb (2.74 ± 0.81 $N \cdot kg^{-1}$; CI: 2.54-3.32) compared with non-dominant limb (2.61 ± 0.72 $N \cdot kg^{-1}$; CI: 2.27-2.95, $p = 0.46$, $d = 0.08$) irrespective of movement. There was no significant ($F = 2.11$, $p = 0.06$) movement *x* interaction.

Time to Peak Vertical Force

Pairwise comparisons demonstrated that time to peak $_{V}$GRF during the temps levé (0.10 ± 0.01 s; CI: 0.10-0.10) was significantly slower compared with the sissonne PDB (0.09 ± 0.01 s; CI: 0.08-0.09, $p < 0.01$, $d = 0.54$). Time to peak $_{V}$GRF during the temps levé, sissonne PDB, and the sissonne (0.10 ± 0.02 s; CI: 0.09-0.11) was significantly slower than the jeté (0.07 ± 0.02 s; CI: 0.06-0.08, $p \leq 0.01$, d = 0.57-0.71), jeté step (0.07 ± 0.02 s; CI: 0.06-0.07, $p \leq 0.01$, d = 0.60-0.73), and the échappé (0.07 ± 0.01 s; CI: 0.06-0.08. $p \leq 0.01$, d = 0.68-0.82). Finally, time to peak $_{V}$GRF during the sissonne was significantly slower compared with the jeté ET (0.09 ± 0.02 s; CI: 0.08-0.10, $p = 0.04$, $d = 0.33$), which was significantly slower than the jeté step ($p = 0.01$, $d = 0.45$) and jeté ($p = 0.02$, $d = 0.41$). The movement *x* limb interaction was also significant ($p = 0.02$), demonstrating that time to peak $_{V}$GRF was significantly quicker in the non-dominant limb compared with the dominant limb during the sissonne PDB (0.08 ± 0.07 s vs 0.09 ± 0.04 s; CI: 0.003-0.01, $p < 0.01$, $d = 0.51$) and temps levé (0.10 ± 0.01 s vs 0.10 ± 0.01 s; CI: 0.004-0.01, $p < 0.01$, $d = 0.44$) respectively

Mean Vertical Loading Rate

Post-hoc comparisons identified that mean $_{V}$LR during the jeté ET (40.52 ± 17.11 $BW \cdot s^{-1}$; CI: 31.17-49.88) compared with the échappé (28.44 ± 13.28 $BW \cdot s^{-1}$; CI: 21.06-35.81, $p = 0.04$, $d = 0.37$). No further significant differences between movements were highlighted ($p \geq 0.05$), and there was no significant movement *x* limb interaction ($F = 1.88$, $p = 0.10$)

Kinematic parameters

Table 4.4 and Figures 4.1-4.3 summarise the influence of movement task and laterality on the kinematic responses in each movement plane. There was a significant main effect identified for movement ($p < 0.01$) for each metric in each movement plane. However, there was no significant ($p \geq 0.22$) main effect for limb identified in any metric within each movement plane. When delineating the post-hoc comparisons following the significant main effects for movement, the values presented in the text under each dependent variable sub-heading, are averaged from the dominant and non-dominant limb. There was no significant movement *x* limb interaction ($p \geq 0.08$) for any kinematic metric.

Table 4.4. 3D kinematic data for the dominant and non-dominant limb during execution of the dance-specific jump landing manoeuvres. Values are mean ± σ.

Sagittal		Jump Landing Manoeuvre						
Variable	**Limb**	**Jeté**	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
Angle at FS (°)	Dominant	29.27 ± 4.78$^{\delta 7}$	31.58 ± 6.97$^{\delta 7}$	29.60 ± 4.19$^{\delta 7}$	29.26 ± 5.44$^{\delta 7}$	31.58 ± 7.06	31.38 ± 7.83$^{\delta 7}$	37.69 ± 4.78$^{\delta 12346}$
	Non-Dominant	28.82 ± 5.52$^{\delta 7}$	32.38 ± 8.22$^{\delta 7}$	29.57 ± 5.23$^{\delta 7}$	29.94 ± 5.06$^{\delta 7}$	33.32 ± 6.36	31.37 ± 7.60$^{\delta 7}$	37.28 ± 5.52$^{\delta 12346}$
Peak angle (°)	Dominant	92.37 ± 5.57$^{\delta 5}$	100.16 ± 5.38$^{\delta 3}$	91.89 ± 2.94$^{\delta 267}$	95.70 ± 5.11$^{\delta 6}$	98.27 ± 6.35$^{\delta 1}$	99.64 ± 4.35$^{\delta 34}$	98.84 ± 5.08$^{\delta 3}$
	Non-Dominant	92.28 ± 5.83$^{\delta 5}$	99.94 ± 5.67$^{\delta 3}$	91.29 ± 4.44$^{\delta 267}$	95.07 ± 5.25$^{\delta 6}$	98.06 ± 6.63$^{\delta 1}$	98.87 ± 5.21$^{\delta 34}$	98.76 ± 5.02$^{\delta 3}$
RROM (°)	Dominant	63.10 ± 6.34$^{\delta 2}$	68.59 ± 6.10$^{\delta 17}$	62.29 ± 3.46	66.44 ± 7.62	66.69 ± 11.31	68.26 ± 7.50$^{\delta 7}$	61.15 ± 9.37$^{\delta 26}$
	Non-Dominant	63.47 ± 5.96$^{\delta 2}$	67.56 ± 5.66$^{\delta 17}$	61.72 ± 4.08	65.13 ± 4.53	64.74 ± 8.64	67.50 ± 7.51$^{\delta 7}$	61.48 ± 8.58$^{\delta 26}$
Coronal								
Angle at FS (°)	Dominant	-2.19 ± 3.18$^{\delta 36}$	-4.06 ± 4.70$^{\delta 36}$	4.55 ± 5.05$^{\delta 1\text{-}7}$	-1.61 ± 5.31$^{\delta 36}$	-2.52 ± 6.87$^{\delta 36}$	-8.37 ± 3.13$^{\delta 13457}$	-0.53 ± 5.17$^{\delta 36}$
	Non-Dominant	-2.64 ± 4.92$^{\delta 36}$	-5.09 ± 6.49$^{\delta 36}$	5.55 ± 4.06$^{\delta 1\text{-}7}$	-1.15 ± 5.21$^{\delta 36}$	-0.80 ± 4.13$^{\delta 36}$	-7.31 ± 4.64$^{\delta 13457}$	-1.45 ± 5.52$^{\delta 36}$
Peak angle (°)	Dominant	-25.78 ± 4.44$^{\delta 3}$	-23.38 ± 3.90$^{\delta 356}$	-11.71 ± 4.85$^{\delta 1\text{-}7}$	-26.34 ± 7.48$^{\delta 3}$	-30.06 ± 5.45$^{\delta 237}$	-27.93 ± 5.82$^{\delta 23}$	-25.30 ± 5.49$^{\delta 35}$
	Non-Dominant	-25.98 ± 7.75$^{\delta 3}$	-22.28 ± 6.36$^{\delta 356}$	-7.88 ± 5.47$^{\delta 1\text{-}7}$	-24.78 ± 7.84$^{\delta 3}$	-28.12 ± 7.44$^{\delta 237}$	-25.27 ± 7.00$^{\delta 23}$	-25.75 ± 8.71$^{\delta 35}$
RROM (°)	Dominant	23.59 ± 4.17$^{\delta 3}$	19.32 ± 4.13$^{\delta 457}$	16.26 ± 3.17$^{\delta 14567}$	24.73 ± 4.55$^{\delta 236}$	27.54 ± 5.86$^{\delta 236}$	19.56 ± 4.77$^{\delta 3457}$	24.77 ± 4.14$^{\delta 236}$
	Non-Dominant	23.34 ± 6.72$^{\delta 3}$	17.19 ± 4.79$^{\delta 457}$	13.43 ± 5.79$^{\delta 14567}$	23.63 ± 5.61$^{\delta 236}$	27.32 ± 6.70$^{\delta 236}$	17.96 ± 6.22$^{\delta 3457}$	24.30 ± 5.80$^{\delta 236}$
Transverse								
Angle at FS (°)	Dominant	-11.15 ± 4.78$^{\delta 3457}$	-10.41 ± 4.87$^{\delta 3457}$	-17.06 ± 6.07$^{\delta 126}$	-15.34 ± 5.45$^{\delta 1267}$	-14.31 ± 5.82$^{\delta 1267}$	-11.36 ± 6.18$^{\delta 3457}$	-19.61 ± 6.12$^{\delta 12456}$
	Non-Dominant	-8.94 ± 4.28$^{\delta 3457}$	-9.06 ± 5.01$^{\delta 3457}$	-17.03 ± 4.07$^{\delta 126}$	-14.55 ± 3.22$^{\delta 1267}$	-14.07 ± 4.28$^{\delta 1267}$	-8.78 ± 3.80$^{\delta 3457}$	-16.17 ± 4.53$^{\delta 12456}$
Peak angle (°)	Dominant	-4.52 ± 2.91$^{\delta 2347}$	-18.09 ± 6.36$^{\delta 1\text{-}7}$	-24.64 ± 3.89$^{\delta 1\text{-}7}$	-6.39 ± 5.02$^{\delta 1\text{-}7}$	-1.73 ± 4.64$^{\delta 23457}$	-3.86 ± 4.28$^{\delta 23467}$	-10.78 ± 5.02$^{\delta 1\text{-}7}$
	Non-Dominant	-2.30 ± 4.81$^{\delta 2347}$	-17.94 ± 6.78$^{\delta 1\text{-}7}$	-26.53 ± 5.43$^{\delta 1\text{-}7}$	-9.08 ± 5.29$^{\delta 1\text{-}7}$	-1.78 ± 5.53$^{\delta 23457}$	-4.75 ± 4.61$^{\delta 23467}$	-8.67 ± 5.91$^{\delta 1\text{-}7}$
RROM (°)	Dominant	6.63 ± 4.62$^{\delta 5}$	7.68 ± 5.45	7.58 ± 3.14$^{\delta 5}$	8.96 ± 5.98$^{\delta 5}$	12.58 ± 5.37$^{\delta 13467}$	7.51 ± 6.79$^{\delta 5}$	8.83 ± 5.04$^{\delta 5}$
	Non-Dominant	6.63 ± 4.99$^{\delta 5}$	8.88 ± 6.73	9.50 ± 3.53$^{\delta 5}$	5.47 ± 3.89$^{\delta 5}$	12.29 ± 5.72$^{\delta 13467}$	4.03 ± 3.89$^{\delta 5}$	7.50 ± 5.31$^{\delta 5}$

FS, footstrike; °, degrees; RROM, relative range of motion; PDB, Pas de Bourres; ET, en Tournant. $^{\delta}$ denotes a significant main effect for movement irrespective of limb, and ^{1234567} signify which movements are significantly different from each other (^{1}) is jeté and (^{7}) is jeté ET

Angle at Footstrike

In the sagittal plane, plantarflexion angle at footstrike during the jeté ET (37.49 ± 8.71°; CI: 32.73-42.24) was significantly lower than all other movements ($p \leq 0.04$, $d = 0.32-0.51$), except for the Sissone PDB (32.45 ± 6.71°; CI: 32.73-42.24, $p = 0.19$). No further significant differences in plantarflexion were identified between the other movements ($p \geq 0.12$) Coronal plane analysis revealed that inversion at footstrike during the échappé (5.05 ± 4.55°; CI: 2.96-7.13) was significantly greater than all other movements ($p \leq 0.02$, $d = 0.48-0.84$). There was significantly more ankle eversion at footstrike during the temps levé (-7.84 ± 3.88°; CI: -9.75- -5.94) compared with all other movements ($p \leq 0.01$, $d = 0.56-0.84$) except for the jeté step (-4.58 ± 5.60°; CI: -7.56- -1.60, $p = 0.82$). No further significant differences between the movements were identified ($p \geq 0.77$). In the transverse plane, foot external rotation during the jeté ET (-17.89 ± 5.33°; CI:-20.19- -15.58) was significantly greater than all other movements ($p \leq 0.04$, $d = 0.29-0.62$) with the exception of échappé (-17.05 ± 5.07°; CI: -19.30- -14.80, $p = 1.00$). External rotation during the échappé, sissonne (-14.91 ± 4.33°; CI: -17.32- -12.70), and sissonne PDB (-14.19 ± 5.05°; CI: -16.75- -11.63 was significantly greater than temps levé (-10.07 ± 4.99°; CI: -12.56- -7.58, $p \leq 0.01$, $d = 0.38-0.57$), jeté step (-9.74 ± 4.94°; CI: -12.41- -7.06, $p \leq 0.008$, $d = 0.41-0.59$) and jeté (-10.05 ± 4.53°; CI: -12.40- -7.69, $p \leq 0.02$, $d = 0.40-0.59$).

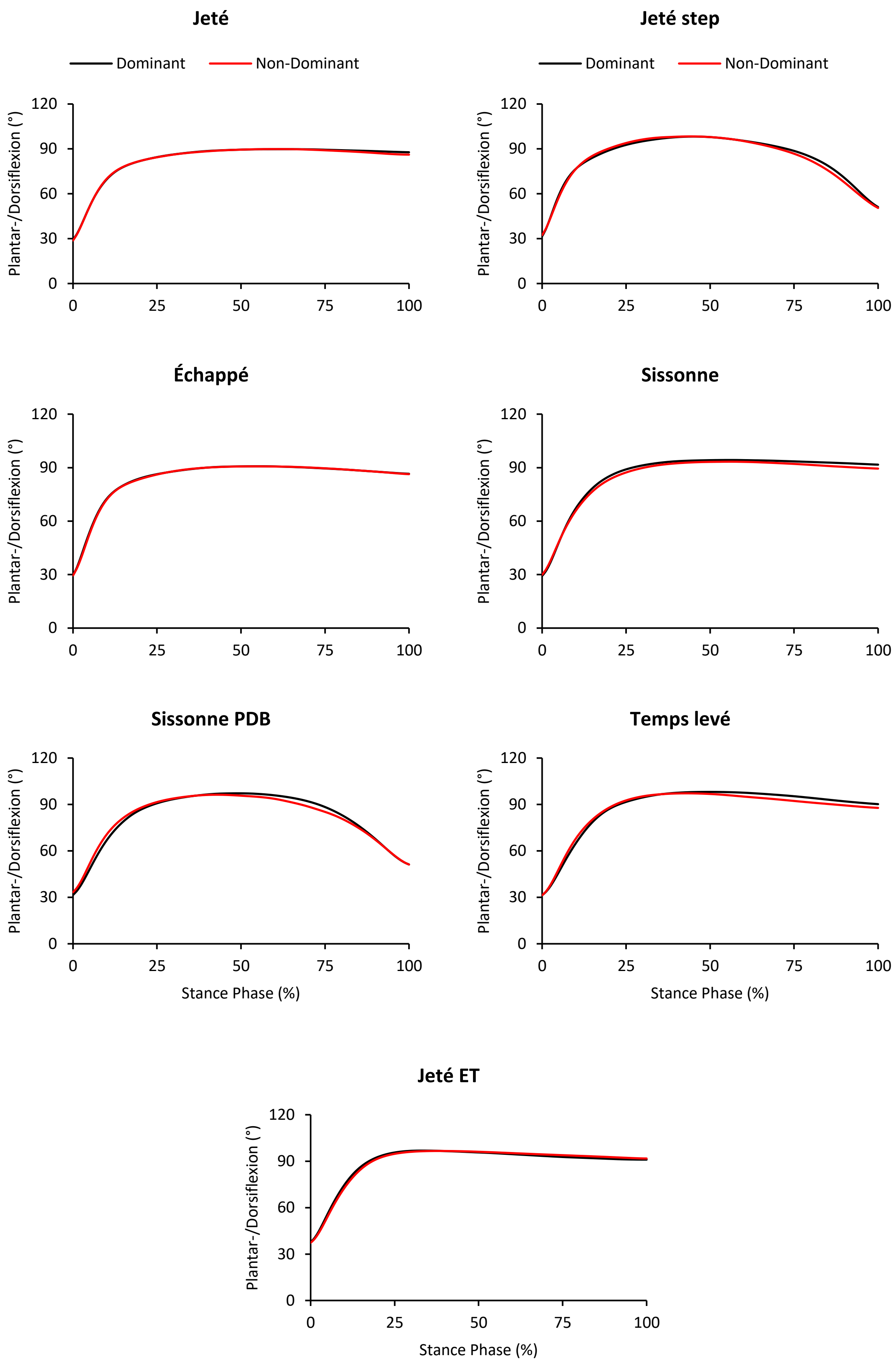

Figure 4.1. Mean sagittal plane ankle joint motion for the dominant (black line) and non-dominant (red line) limb for each jump landing manoeuvre.

Peak Angle

Peak dorsiflexion angle during the échappé (91.59 ± 3.69°; CI: 89.70-83.48) and jeté (92.33 ± 5.70°; CI: 89.12-95.53) was significantly lower than the jeté ET (98.80 ± 5.05°; CI: 96.15-101.45, $p \leq 0.01$, $d = 0.47$-0.59), temps levé (99.25 ± 4.78°; CI: 96.79-101.72, $p = 0.01$, $d = 0.55$-0.67) and jeté step (100.05 ± 5.52°; CI: 97.00-103.10, $p < 0.01$, $d = 0.57$-0.67). In addition, peak dorsiflexion was significantly lower during the jeté than the sissonne PDB (98.17 ± 6.49°; CI: 94.64-101.69, $p = 0.01$, $d = 0.43$), and, during the sissonne (95.39 ± 5.18°; CI: 92.51-98.27) compared with the temps levé ($p = 0.02$, $d = 0.36$). In the coronal plane, peak eversion during the échappé (-9.80 ± 5.16°; CI; -12.49- -7.10) was significantly lower than all other movements ($p < 0.01$, $d = 0.78$-0.86). Further, peak eversion was significantly greater during the sissonne PDB (-29.09 ± 6.44°; CI: -32.51- -25.67) compared with the jeté ET (-25.53 ± 7.10°; CI: -29.15- -21.90, $p = 0.03$, $d = 0.25$) and jeté step (-22.83 ± 5.13°; CI: -25.47- -20.19, $p < 0.01$, $d = 0.47$). The temps levé (-26.60 ± 6.41°; CI: -29.82- -23.37) had significantly greater peak ankle eversion compared with jeté step ($p = 0.02$, $d = 0.31$). Analysis of the transverse plane highlighted that peak external rotation during the échappé (-25.58 ± 4.67°; CI: -27.96- -23.21) was significantly greater than all other movements ($p \leq 0.01$, $d = 0.55$-0.93). Further significant ($p \leq 0.03$) differences between all other movements were identified except for between the jeté (-3.41 ± 3.86°; CI: -4.88- -1.95) and sissonne PDB (-1.76 ± 5.09°; CI: -3.41- -0.11, $p = 0.95$) and temps levé (-4.30 ± 4.45°; CI: -5.95- -2.66, $p = 1.00$), and, between the sissonne PDB and temps levé ($p = 0.18$).

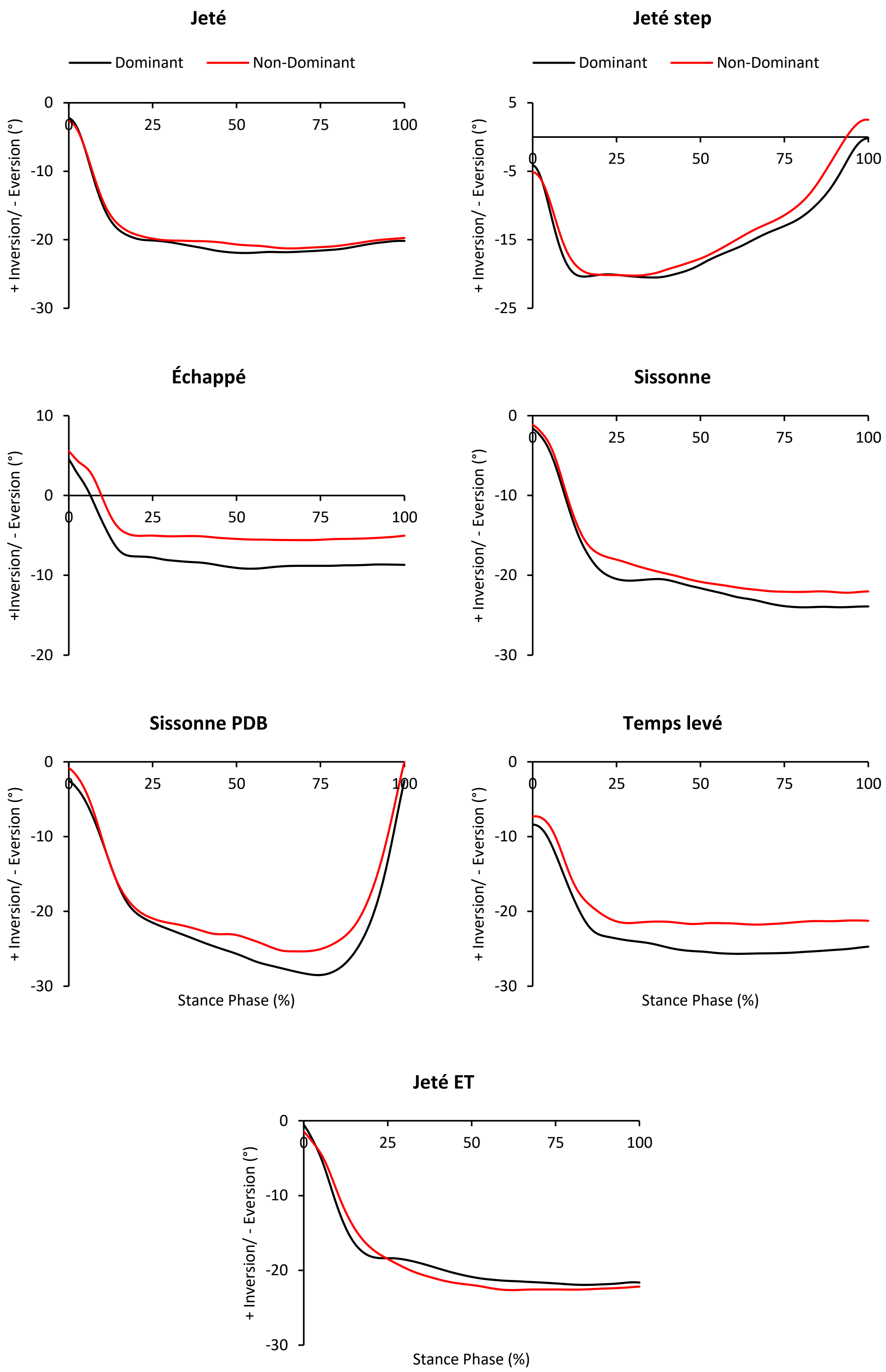

Figure 4.2. Mean coronal plane ankle joint motion for the dominant (black line) and non-dominant (red line) limbs for each jump landing manoeuvre.

Relative Range of Motion

Relative dorsiflexion range of motion during the jeté step (68.07 ± 7.23°; CI: 64.18-71.96) was significantly greater than the jeté (63.29 ± 6.1°; CI: 59.90-66.67, $p < 0.01$, $d = 0.34$) and jeté ET (61.32 ± 8.98°; CI: 56.26-66.37, $p = 0.01$, $d = 0.38$). Further, the temps levé (67.88 ± 7.51 ; CI: 63.85-71.91) had significantly greater relative range of motion compared with the jeté ET ($p = 0.01$, $d = 0.37$). In the coronal plane, relative eversion range of motion was significantly lower in the échappé (14.84 ± 4.48°; CI: 17.25-12.44) compared with all other movements ($p \leq 0.02$) except for the jeté step (18.25 ± 4.64°; CI: 20.69-15.81, $p = 1.00$). The temps levé (18.75 ± 5.50°; CI: 21.26-16.25) and jeté step had a significantly lower relative range of motion compared with sissonne (24.18 ± 5.08°; CI: 26.49-21.87, $p < 0.01$, $d = 0.46$-0.52), jeté ET (24.53 ± 4.97°; CI: 26.97-22.09, $p \leq 0.04$, $d = 0.48$-0.55), and sissonne PDB (27.43 ± 6.28°; CI: 30.35- 24.51, $p < 0.01$, $d = 0.59$-0.64). Transverse plane analysis highlighted significantly lower relative range of motion in the sissonne PDB (12.43 ± 5.55°; CI: 10.55-14.31) compared with all other movements ($p \leq 0.04$, $d = 0.43$-0.55) except for the jeté step (8.28 ± 6.09°; CI: 4.87-11.67, $p = 1.00$). No further significant differences ($p \geq 0.24$) were identified.

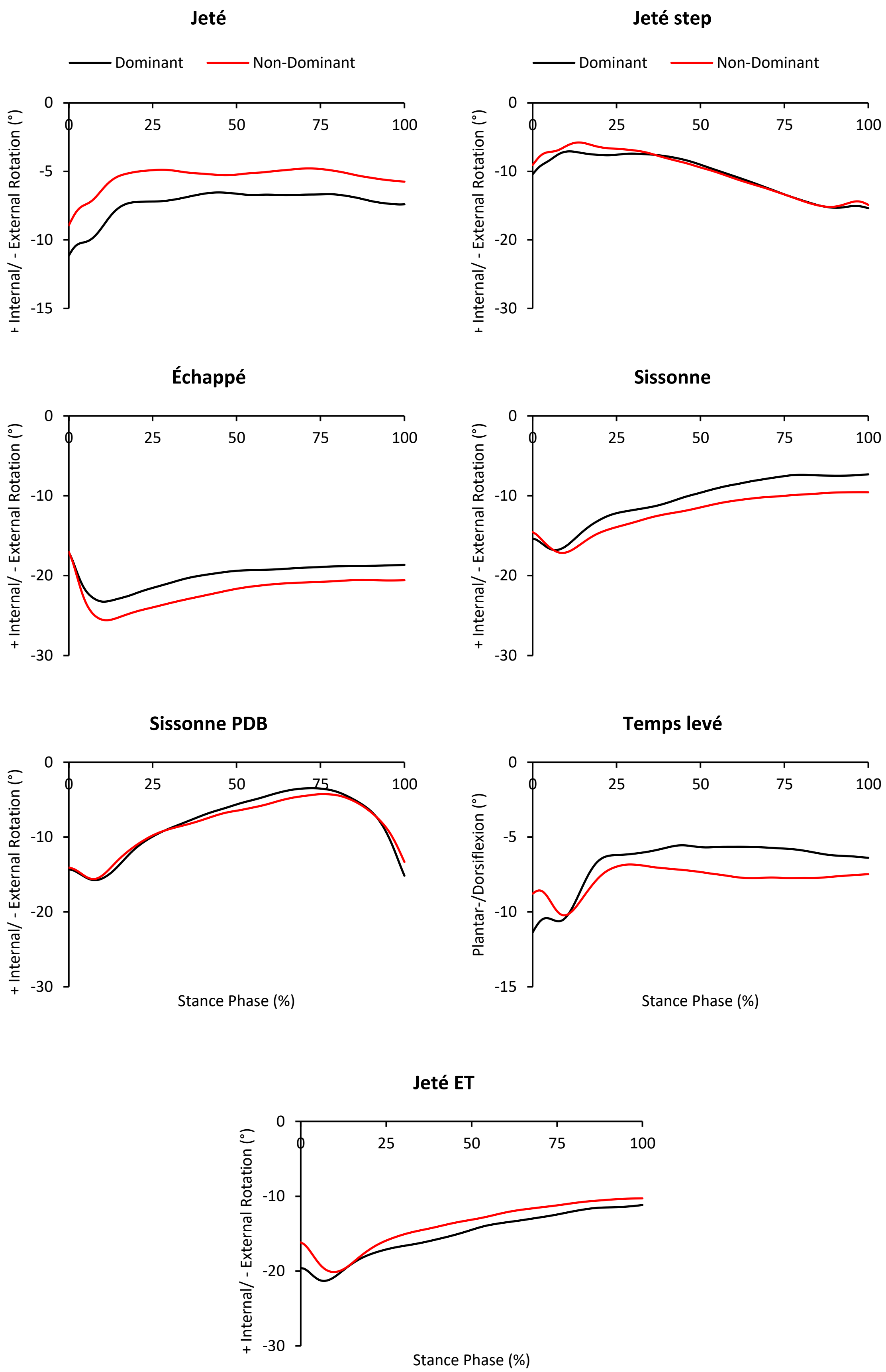

Figure 4.3. Mean transverse plane ankle joint motion for the dominant (black line) and non-dominant (red line) limbs for each jump landing manoeuvre.

4.4 Discussion

The aim of the current study was to investigate the ankle joint kinematic and kinetic responses to a battery of ballet-specific jump landing manoeuvres of varying planar demand. Currently, there is a dearth of information on the biomechanics allied to multiple variants of ballet-specific movement. The battery of tasks used in the current investigation provides methodological rationale, and focusing analyses at the ankle joint corresponds with injury epidemiology observations.

The tasks incorporated in this study varied in planar demand and required both single- and double-leg execution. In addition, the selected techniques differentiated between landings to achieve a cessation of momentum, and, landings which act as transitions into the next movement. As anticipated, and reflecting the range in technical characteristics of the movement battery, there was a significant main effect highlighted for movement in each kinetic parameter. Peak vertical GRFs ranged between 1.84-3.09 body weights for the échappé and jeté ET respectively. Direct comparisons with the literature are limited by a lack of studies and should be treated with caution, since the biomechanical response is likely to be influenced by the experimental methodology and the participants, with previous literature considering pre-professional (Kulig, Fietzer and Popovich, 2011; Fietzer, Chang and Kulig, 2012) or professional (Jarvis and Kulig, 2016) ballet dancers. The échappé demonstrated similar $_V$GRF magnitudes to those reported in a comparable cohort of university dancers (Peng et al., 2015), suggesting corroborative findings when comparing ballet dancers of similar ability within the same movements. However, the vertical GRFs during the landing phase of the jeté were lower than those reported during a different but albeit analogous movement in the saut de chat, in dancers enrolled on elite pre-professional programmes (Jarvis and Kulig, 2016; Kulig, Fietzer and Popovich, 2011). Comparing the results of studies with different skill levels of performers is difficult, especially when the technical descriptor (movement velocity, jump height) vary between studies, which likely influences the magnitude of impact forces (Grabowski and Kram, 2008).

The transitional landing component incorporated into specific manoeuvres of the current study better simulate the locomotor patterns allied to ballet performance, with dancers often performing multiple techniques in rapid succession. However, analyses revealed that the transitional element of selected tasks had no significant effect on resultant GRF metrics. Specifically, vertical GRFs between the jeté and jeté step, and between the sissonne and sissonne PDB were marginally (2.4% and 6.5% respectively) lower. The improved ecological validity of the current investigation contrasts with existing methodologies in which peak $_V$GRF in female ballet dancers have been quantified during single-leg drop landings (Orishimo et al., 2009), or, an isolated ballet task (Kulig, Fietzer and Popovich, 2011; Lee et al., 2012; Peng et al., 2015; Jarvis and Kulig, 2016). Drop landing tasks negate the multi-planar characteristics of ballet, whilst analysing a single manoeuvre limits the level of biomechanical understanding regarding the mechanics of ballet performance. Hence, statistical inferences should be made with caution when examining landing kinetics in female ballet dancers given the variation in the task of analysis used. Valid and meaningful comparisons may only be achievable between studies with uniform methodological designs (Russell, 2013). With a dearth of biomechanics information available in ballet dancers using a functionally relevant testing battery, further research on the mechanical rigours of discrete ballet techniques is warranted.

A second important finding was the significant effect for movement on resultant joint kinematics. Sagittal plane joint displacement at footstrike ranged between 53° (jeté ET) and 61° (jeté) of plantarflexion, whilst peak dorsiflexion angles over the stance phase ranged between 1.5° (échappé) to 10° (jeté step). Overall sagittal plane ranges of motion were 61° (jeté ET) to 68° (jeté step). Data from goniometry analyses has highlighted that female university and professional ballet dancers can assume ankle joint excursions of 74° and 35° in plantar- and dorsiflexion respectively, in active weight bearing conditions (Russell et al., 2010). These observations suggest that ballet dancers perform ballet-specific techniques within their physical constraints, but the pointe technique and the repeated exposure to high magnitude multi-planar loading may partially explain the high incidence of ankle pathology in this population.

Also evident in the kinematic data was a loading issue in the coronal plane, with data demonstrating increasing ankle joint eversion over the duration of stance phase, and with more marked ankle eversion evident in the land and transition tasks. Relative eversion range of motion varied between 15° for the échappé to 27° for the sissonne PDB. The stop and hold requirement of the sissonne elicited approximately 12.5% lower peak ankle eversion compared with the transitional sissonne PDB. The greater eversion displacement evident in the sissonne PDB may be symptomatic of pre-anticipatory movement towards a transition into a subsequent task. Though not evident in the kinetic measures, the current data suggests that incorporating a transitional element into specific manoeuvres, which more closely simulates true performance motion, highlights altered joint displacements compared with stable landing techniques. With ballet dancers often performing successive techniques with a rapid landing transition, the current methodological design has greater specificity to ballet, and the kinematic findings may have important implications for future biomechanical assessments of ballet movement. The dorsiflexed and everted ankle configuration is quite different to the typical mechanism of ankle sprain injury described as plantarflexed and inverted (Fousekis, Tsepis and Vagenas, 2012). This observation is particularly evident in the tasks that require a land and hold technique (jeté, échappé, sissonne, temps levé, jeté ET). However, in the movements that contain a transition element (jeté step, sissonne PDB), and more closely replicate typical performance motion, plantarflexion and a shift towards inversion are present towards the latter stages of the stance phase, which may have implications for injury risk. Injury epidemiology often highlights a gross injury location without specifying explicit diagnoses, and the influence of movement task on ankle joint loading and injury risk warrants further attention.

In acknowledgment of the asymmetric movement profile of ballet, the current kinetic and kinematic responses comprised a bilateral consideration. Adopting a particular leg to perform key intricacies, commonly termed limb dominance, may not be as apparent in ballet as with other sports such as soccer (Fousekis, Tsepis and Vagenas, 2012; Daneshjoo et al., 2013). Teaching/rehearsal classes are assumed to prescribe exercises and movements that require equal contribution from both limbs, and therefore, inter-limb balance is implied (Farrar-Baker and Wilmerding, 2006). However, dancers may have a preferential limb in which to 'push off', jump and/or land (Murphy, Connolly and Beynnon, 2003). The analysis of bilateral mechanical response in ballet dancers is important to quantify training adaptations, and to inform injury risk resulting from increased mechanical loading to a particular side (Croisier et al., 2008; Fousekis, Tsepis and Vagenas, 2012). This suggestion is augmented in consideration of the 56% of jump landings in a 200-jump routine comprising a unilateral landing component (Liederbach et al., 2006).

With the exception of peak vertical force, there were no significant between-limb differences across a range of kinematic and kinetic parameters. Furthermore, the lack of bilateral asymmetry was common to all movements, irrespective of task demand. Mertz and Docherty (2012), also reported that 'leg preference' during jump landing tasks failed to influence resultant biomechanical parameters, whilst the same observation has been demonstrated during non-jumping tasks such as the elevé (Abraham et al., 2018). Not only is biomechanical symmetry an important contributor towards aesthetics, and thus performance success, it mitigates against injury potential due to increased loading to a particular limb (Zifchock et al., 2008). Consequently, the clinical ballet medicine profession can potentially conclude that limb dominance does not influence the manner in which university-level dancers manage the ground reaction forces allied to a range of performance-specific jump landing tasks. The bilateral symmetry amongst selected kinematic and kinetic measures evident in this population, who despite their training status and dance experience are not elite or professional by definition, may be indicative of early exposure to ballet movement with emphasis on limb control during training (Bronner and Ojofeitimi, 2006). These strategies may facilitate beneficial adaptations towards the development of bilateral symmetry, with clinical implications apparent in a reduced susceptibility to injury.

In the present study, kinetic responses and ankle joint biomechanics were quantified during anticipated, isolated ballet-specific movements, performed discretely and by non-elite dancers. Generalisations beyond these specific research design elements should be treated with caution, particularly as interactions are likely in the complexity of skill, and level of performer. Performing these highly controlled jump landing tasks in this environment may not appropriately simulate typical performance , in which multiple, consecutive elements are executed, or, truly reflect the mechanism of injury (Lee et al., 2012). Participants in the current study completed the jump landing battery in a non-fatigued state. With the majority of injuries in amateur female ballet dancers attributed to overuse (Smith et al., 2015), and fatigue proposed to alter ankle joint kinematics (Weinhandl, Smith and Dugan, 2011), testing in pre- and post-fatigue conditions may have better reflected true performance rigours and yielded different kinematic and kinetic findings towards understanding the occurrence of injury. Therefore, investigating the influence of fatigue on bilateral lower limb biomechanics across genders and skill level, is an area with scope for subsequent research.

4.5 Conclusion

This study represents the first to investigate the kinetic and ankle joint kinematic responses to a battery of ballet-specific movements in female ballet dancers, with the tasks designed to differentiate between linear and rotational landings, and between stop and hold versus transitional landings. The battery of movement tasks was sufficiently sensitive to highlight a main effect for movement type. Kinetic data showed clear differences in the mechanical rigours of ballet tasks, but that a transitional landing component fails to reduce resultant vertical GRFs. Kinematic observations indicated a transition effect towards greater ankle eversion and inversion over the stance phase. There was no evidence of bilateral asymmetry in ankle kinematics, peak $_V$GRF or in mean $_V$LR, which may be indicative of appropriate training practices that facilitate equal biomechanical development in both limbs from an early age. Data from the current study highlighted kinetic and kinematic differences in ballet-specific movement tasks that are common constructs of practice, choreography and performance. The current study provides important information on the mechanical characteristics of ballet locomotion, and may inform load management and athlete monitoring and screening practices in ballet.

Chapter 5. Study two: Electromyographic Responses to a Ballet-specific Movement Battery

5.1 Introduction

The jump landing testing battery used in study 1 was sufficiently sensitive to differentiate the kinetic demands and ankle joint kinematic responses allied to ballet-specific movements typical of a routine. Regulation of these high impact and complex techniques requires optimal lower limb kinematic control, to which the neuromuscular system plays a vital role (Wilson, Lim and Kwon, 2004). Intuitively, a similar between-movement response would be observed at the muscular level reflecting the variation in task characteristics. There is currently limited information available on the lower limb electromyographic (EMG) responses to ballet movement, especially the jump landing tasks inherent to performance. Consideration of the muscular strategy governing landing mechanics may provide important aetiological data towards the high prevalence of ankle injury in ballet dancers (Smith et al., 2015).

The neuromuscular system is central to the mechanisms that enable the body to move efficiently and aesthetically (Thullier and Moufti, 2004). It also has a key role in providing joint stability during landing, and mobility when transitioning into a subsequent movement (Brockett and Chapman, 2016). The pointe technique in ballet kinaesthetics, and the eversion/inversion alignment common to ballet-specific jump landing tasks as demonstrated in study 1, provides a rationale for the selection of muscles used in EMG analysis. Globally, the peroneal muscles facilitate mediolateral stability to resist the eversion/inversion mechanism, whilst the tibialis anterior and the lateral gastrocnemius govern dorsi- and plantarflexion motion respectively (Bavdek, 2018). Hence, concerted activation of these muscles is crucial towards ankle joint kinematics when absorbing the ground reaction forces (GRFs) generated during ballet movement, with important implications towards a reduced injury risk.

EMG has featured in numerous ballet studies, primarily to quantify the muscular contributors to discrete techniques such as the plié (Couillandre, Lewton-Brain and Portero, 2008; Lin 2015), relevé (Massó et al., 2004; de Bartolomeo et al., 2007; Lin et al., 2016), and the pirouette (Zaferiou et al., 2017). Whilst this information is valuable for the physical and technical development of dancers, the aesthetic characteristics of these floor exercises underlines an evident performance enhancement focus. The neuromuscular responses to jump landing tasks, which are regarded as the primary locomotor contributors to injury (Fousekis, Tsepis and Vagenas, 2012), have largely been ignored within existing methodological designs. Where relevant information is available, analyses have incorporated a clinical drop landing task (Turner et al., 2018), or a single ballet-specific movement (Lin et al., 2012). Previous studies have therefore lacked specificity, limiting an understanding on the neuromuscular strategy used during a range of ballet-specific jump landing tasks with varying kinematic profiles.

The aim of current study therefore was to evaluate the neuromuscular responses of the lower limb to the same testing battery used in study 1. Study 1 highlighted a task-dependent response in kinetics and ankle joint kinematics, which likely influences the EMG strategy used. In addition, the mechanical responses provided insight into the functional kinesiology requirements of ballet movement, supporting the choice of musculature comprising the subsequent EMG analysis. In consideration of the requirement for equal limb contribution during ballet performance, bilateral symmetry in EMG response is also considered.

Experimental research question:

1) Do variations in ballet task landing instruction ('hold' vs 'transition') elicit differences in the magnitudes of EMG response.

5.2 Methods

Participants

The participants used during study 2 were the same female cohort that completed study 1 (please refer to Section 4.2 for further information).

Experimental Procedures

The experimental protocol regarding the warm-up, exercise familiarisation and the experimental trials were also identical to those outlined in Chapter 4.2. Participants were required to wear shorts, to enable direct contact between the EMG electrode and skin for analysis.

Data Collection

A surface EMG system (Noraxon, Noraxon USA inc, Arizona, USA) was used to record neuromuscular activity, and sampled data at a frequency of 1500 Hz. Appropriate skin surface preparation for each participant was completed, and involved shaving (standard razor), abrading (rough brushing of the skin with paper towel) and cleansing (70% isopropyl alcohol wipe) the desired area to reduce impedance at the skin-electrode interface. Disposable, self-adhesive, Ag/AgCL solid gel dual electrodes (Noraxon, Noraxon USA inc, Arizona, USA) with an inter-electrode distance of 2 cm, were placed on the skin in orientation of the muscle fibre alignment in accordance with the guidelines prescribed by SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles). A two snap lead connected the electrode to a small telemetric module in close proximity, allowing signals to be transmitted telemetrically to a nearby desktop DTS receiver. Surface EMG was applied bilaterally to the Tibialis Anterior (TA), Peroneus Longus (PL) and Lateral Gastrocnemius (LG). Due to their narrow arrangement, the EMG traces corresponding to TA and PL were examined for potential cross-talk. This procedure required each participant to assume foot positions of ankle plantar-/dorsiflexion and eversion/inversion, with resultant EMG amplitudes inspected for similarities and differences. Initially, an electrode was applied solely to the LG to examine the neuromuscular response to a series of heel raises. A second electrode was then positioned on the PL, with EMG responses first assessed using the heel raise. Satisfied that the heel raises elicited different responses between muscles (lower activation of PL), active weight-bearing ankle eversion and inversion positions were assumed to validate initial observations. On this occasion, PL activity was greater than LG, reflecting the functional responsibilities inherent to each muscle. The third electrode was attached to the TA, with active ankle dorsiflexion assumed to examine resultant activations, and to further verify that cross-talk was not present.

Data Processing

During the experimental trials, corresponding EMG signals were recorded and interfaced synchronously with the force plate using QTM software. Resultant analog signals were converted into C3D format and exported to Visual 3-D for further analysis. The EMG data was initially bandpass filtered using 20 Hz (high-pass) and 450 Hz (low-pass) cut-offs to minimise contaminated data in the form of movement artefact and high frequency noise. Data was then full-wave rectified, and smoothed using the Root Mean Squared (RMS) method over a 75 ms time constant. Resultant EMG envelopes were normalised to the stance phase and were subsequently analysed to obtain mean and peak EMG from the amplitude domain.

Statistical Analysis

IBM SPSS Statistics V25.0 software (IBM, Armonk, New York, USA) was used to analyse the data. Descriptive statistics (mean ± σ) are presented for each EMG parameter. histograms, Q-Q plots and a Shapiro-Wilk statistic were employed to quantify the distribution of the data. With the data normality assumptions verified, separate repeated measures 7 *x* 2 (movement *x* muscle *x* limb) ANOVAs were used to determine the significance of the outcome measures across the movement battery for each muscle. Where a statistically significant main effect and/or interaction occurred, post-hoc pairwise comparisons with a Bonferroni correction factor determined which variables were significantly different from each other. 95% Confidence Intervals (CIs) and Cohen's *d* effect sizes (small; 0.20-0.49, moderate; 0.50-0.79, large; > 0.80) are also presented (Cohen, 1988). A $p < 0.05$ value was used as the statistical significance threshold.

5.3 Results

Table 5.1 and Figures 5.1-5.3 summarise the influence of ballet-specific movement task and laterality on resultant EMG responses. Significant main effects and interactions involving movement and muscle were identified. However, there was no significant evidence for the limb variable.

Table 5.1. Amplitude and frequency domain EMG parameters of selected bilateral muscles during the jump landing testing battery. Corresponding values are to the nearest whole µV and are expressed as mean ± σ.

Muscle Parameter	**Limb**	**Jump-Landing Manoeuvre**						
		Jeté^{*12}	**Jeté Step**^{*1}	**Échappé**^{*1}	**Sissonne**	**Sissonne PDB**	**Temps Levé**	**Jeté ET**
LG								
Mean EMG	Dominant	71 ± 31$^{\delta 2}$	99 ± 36^{1346}	53 ± 22$^{\delta 25}$	53 ± 21$^{\delta 25}$	93 ± 36$^{\delta 346}$	51 ± 19$^{\delta 25}$	65 ± 50$^{\delta}$
(µV)	Non-Dominant	76 ± 58$^{\delta 2}$	106 ± 58^{1346}	51 ± 33$^{\delta 25}$	52 ± 24^{25}	86 ± 35$^{\delta 346}$	59 ± 26$^{\delta 25}$	66 ± 30$^{\delta}$
Peak EMG	Dominant	193 ± 57$^{\delta}$	331 ± 143^{347}	149 ± 62$^{\delta 25}$	149 ± 84$^{\delta 2}$	245 ± 77$^{\delta 3}$	149 ± 54$^{\delta}$	172 ± 102$^{\delta 2}$
(µV)	Non-Dominant	218 ± 103$^{\delta}$	280 ± 144^{347}	146 ± 68$^{\delta 25}$	130 ± 50$^{\delta 2}$	243 ± 95$^{\delta 3}$	176 ± 96$^{\delta}$	176 ± 74$^{\delta 2}$
PL								
Mean EMG	Dominant	169 ± 69$^{\gamma 37}$	162 ± 73$^{\gamma}$	119 ± 48$^{\gamma 1}$	161 ± 71	159 ± 77	126 ± 30	108 ± 41^{1}
(µV)	Non-Dominant	190 ± 100$^{\gamma 37}$	181 ± 86$^{\gamma}$	114 ± 58$^{\gamma 1}$	150 ± 65	184 ± 89	181 ± 79	144 ± 72^{1}
Peak EMG	Dominant	374 ± 94^{7}	479 ± 241	352 ± 225	333 ± 148	435 ± 193$^{\gamma}$	289 ± 88	237 ± 94$^{\gamma 1}$
(µV)	Non-Dominant	474 ± 229^{7}	495 ± 254	346 ± 166	345 ± 190	519 ± 276$^{\gamma}$	423 ± 160	368 ± 165$^{\gamma 1}$
TA								
Mean EMG	Dominant	226 ± 108$^{\gamma 25}$	134 ± 49$^{\gamma 13467}$	202 ± 88$^{\gamma 2}$	228 ± 121^{25}	140 ± 56^{147}	231 ± 141^{2}	276 ± 131^{25}
(µV)	Non-Dominant	234 ± 125$^{\gamma 25}$	103 ± 54$^{\gamma 13467}$	167 ± 86$^{\gamma 2}$	212 ± 127^{25}	123 ± 62^{147}	175 ± 83^{2}	262 ± 148^{25}
Peak EMG	Dominant	466 ± 237	423 ± 205^{7}	416 ± 121	464 ± 218	370 ± 155$^{\gamma 7}$	466 ± 238	546 ± 250$^{\gamma 25}$
(µV)	Non-Dominant	420 ± 217	291 ± 171^{7}	389 ± 178	437 ± 201	328 ± 181$^{\gamma 7}$	399 ± 215	517 ± 258$^{\gamma 25}$

LG, Lateral Gastrocnemius, PL, Peroneus Longus; TA, Tibialis Anterior; µV, Microvolts. * denotes a significant main effect for movement, irrespective of muscle or limb. $^{\delta}$ denotes that LG is significantly lower than PL and TA. $^{\gamma}$ denotes a significant difference between TA and PL. ^{1234567} signifies which movements were significant different from each other (^{1}) is jeté and (^{7}) is jeté ET.

Mean EMG

Lateral Gastrocnemius

The repeated measures ANOVA revealed a significant main effect for movement (F = 9.14, p < 0.01). Post-hoc pairwise comparisons highlighted that LG activation was significantly higher in during the jeté step (102 ± 47 µV; CI: 80-128 µV) compared with the jeté (73 ± 45 µV; CI: 53-94 µV, p = 0.02, *d* = 0.30), échappé (52 ± 27 µV; CI: 41-64 µV, p < 0.01, *d* = 0.55), sissonne (53 ± 23 µV; CI: 42-64 µV, p = 0.04, *d* = 0.56), and the temps levé (55 ± 22 µV; CI: 46-64 µV, p = 0.03, *d* = 0.54). In addition, LG mean EMG was significantly greater during the sissonne PDB (90 ± 35 µV; CI: 73-106 µV) compared with the échappé (p < 0.01, *d* = 0.51), sissonne (p = 0.01, *d* = 0.53), and the temps levé (p = 0.03, *d* = 0.50). There was no significant main effect identified for limb (F = 0.11, p = 0.75), nor a significant movement *x* limb interaction (F = 0.29, p = 0.94).

Peroneus Longus

A significant main effect for movement (F = 4.32, p < 0.01) was identified. Post-hoc analyses revealed that mean EMG was significantly lower during the échappé (117 ± 53 µV; CI: 95-139 µV) than the jeté (180 ± 84 µV; CI: 146-213, p = 0.01, *d* = 0.41) and jeté step (171 ± 79 µV; CI: 134-208 µV, p = 0.04, *d* = 0.38), and during the jeté ET (126 ± 56 µV; CI: 98-153 µV) compared with jeté (p < 0.01, *d* = 0.35). There was no evidence of a significant main effect for limb (F = 1.18, p = 0.30), and no significant movement *x* limb interaction (F = 1.72, p = 0.13).

Tibialis Anterior

ANOVA highlighted a significant main effect for movement (F = !2.39, p < 0.01), with post-hoc comparisons demonstrating that mean EMG response was significantly lower during the jeté step (119 ± 51 µV; CI: 94-143 µV) compared with all other movements (p ≤ 0.03, *d* = 0.42-0.58), except for the sissonne PDB (132 ± 59 µV; CI: 109-155 µV, p = 1.00). Further, the sissonne PDB had significantly lower mean EMG responses compared with the jeté (230 ± 116 µV; CI: 171-289 µV, p = 0.03, *d* = 0.47), sissonne (220 ± 124 µV; CI: 158-282 µV, p = 0.04, *d* = 0.41), and the jeté ET (269 ± 140 µV; CI: 198-339 µV, p = 0.01, *d* = 0.54). There was no significant main effect for limb (F = 1.43, p = 0.25), nor a significant movement *x* limb interaction (F = 0.76, p = 0.58)

Peak EMG

Lateral Gastrocnemius

The repeated measures ANOVA identified a significant main effect for movement (F = 12.79, p < 0.01). Post-hoc analysis revealed that peak EMG was significantly higher during the jeté step (306 ± 144 µV; CI: 244-376 µV) compared with the échappé (148 ± 65 µV; CI: 18-178 µV, p < 0.03, *d* = 0.58), sissonne (139 ± 67 µV; CI: 105-174 µV, p = 0.01, *d* = 0.60), and the jeté ET (174 ± 88 µV; CI: 131-217 µV, p = 0.02, *d* = 0.48). Moreover, peak EMG response was significantly higher during the sissonne PDB (244 ± 86 µV; CI: 208-280 µV) compared with the échappé (p < 0.01, *d* = 0.55) and the sissonne (p = 0.01, *d* = 0.56). There was no significant main effect identified for limb (F = 0.05, p = 0.83), and no significant movement *x* limb interaction (F = 0.99, p = 0.44).

Peroneus Longus

A significant main effect for movement (F = 5.27, p < 0.01) was highlighted, with post-hoc analysis revealing that peak EMG was significantly greater during the jeté (424 ± 161 µV; CI: 356-492 µV) compared with the jeté ET (302 ± 130 µV; CI: 235-371 µV, p = 0.01, *d* = 0.38), but no other significant (p ≥ 0.53) between-movement differences were found. There was no significant main effect for limb (F = 2.08, p = 0.17), nor a significant movement *x* limb interaction (F = 1.42, p = 0.22).

Tibialis Anterior

ANOVA yielded a significant main effect for movement (F = 4.73, p < 0.01), with post-hoc pairwise comparisons demonstrating that peak EMG was significantly greater during the jeté ET (532 ± 254 µV; CI: 400-663 µV) compared with the jeté step (357 ± 188 µV; CI: 267-447 µV, p = 0.04, *d* = 0.36) and sissonne PDB (349 ± 168 µV; CI: 275-423 µV, p = 0.03, *d* = 0.39). There was no significant main effect identified for limb (F = 1.41, p = 0.26), and no significant movement *x* limb interaction (F = 0.66, p = 0.68).

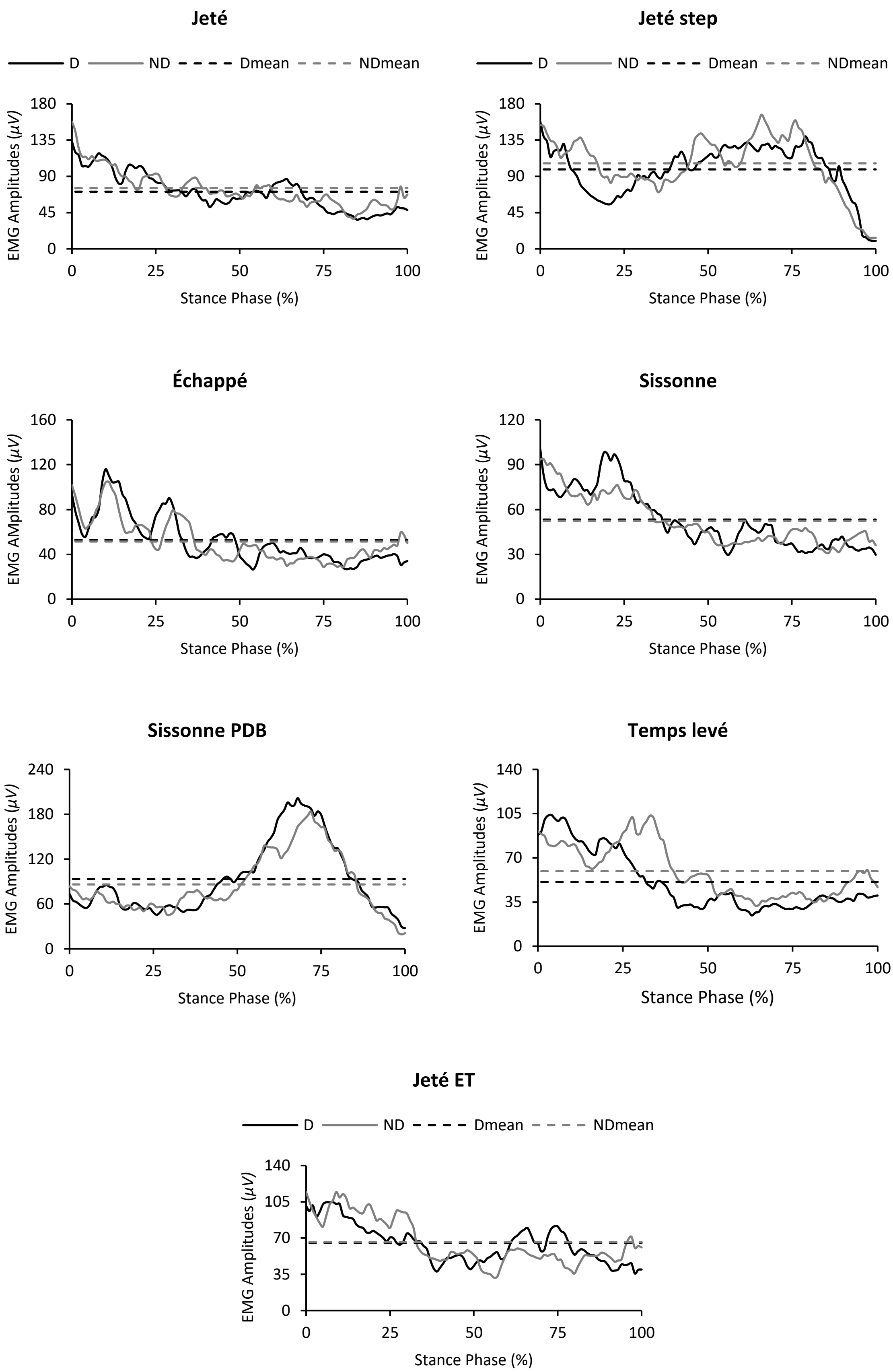

Figure 5.1. Lateral gastrocnemius RMS EMG amplitudes over duration of stance for both the dominant (black solid line) and non-dominant (grey solid line) limb, representing data for all participants and trials. Mean EMG data is presented as dashed lines; dominant (black), non-dominant (grey) limb.

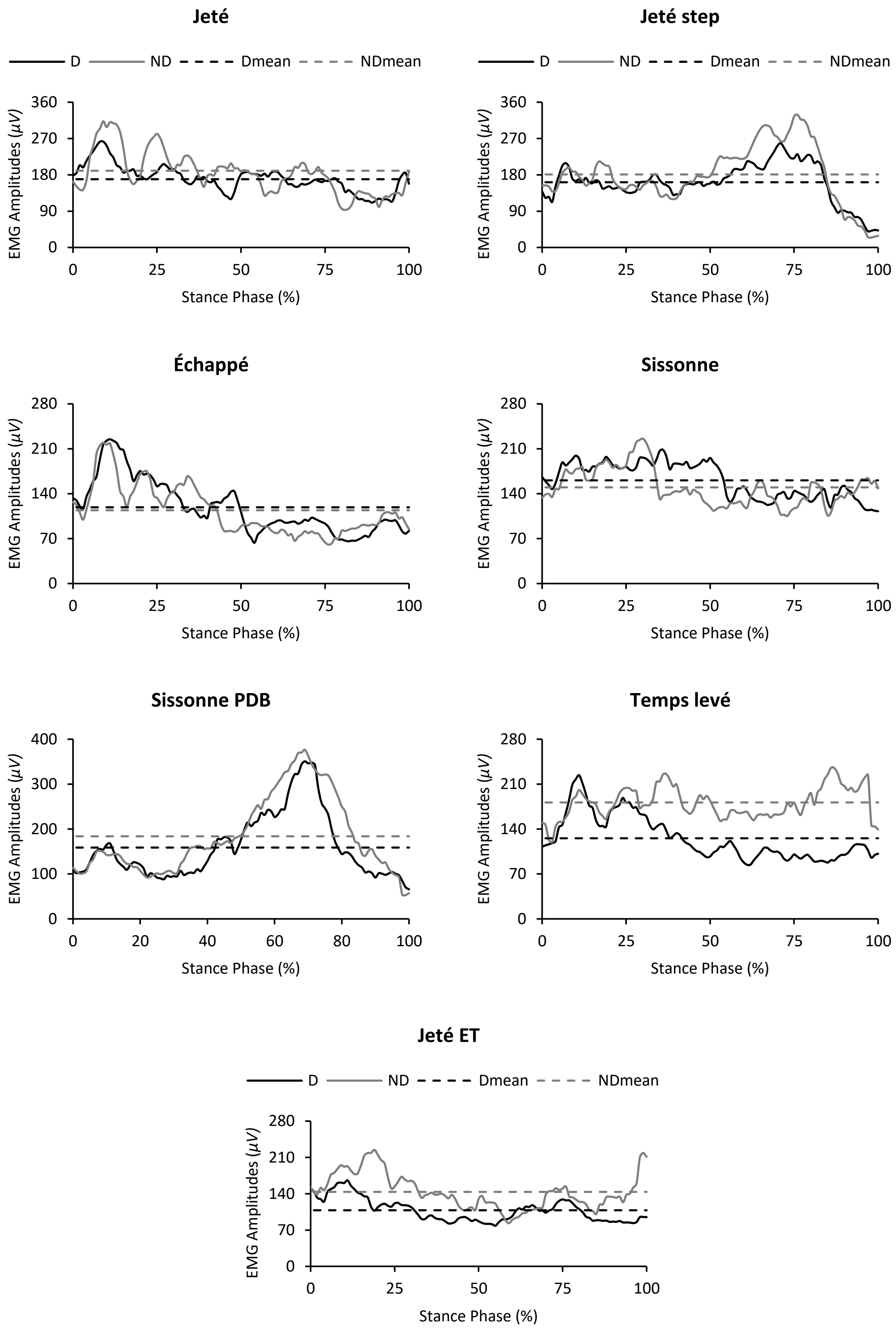

Figure 5.2. Peroneus longus RMS EMG amplitudes over duration of stance for both the dominant (black solid line) and non-dominant (grey solid line) limb, representing data for all participants and trials. Mean EMG data is presented as dashed lines; dominant (black), non-dominant (grey) limb.

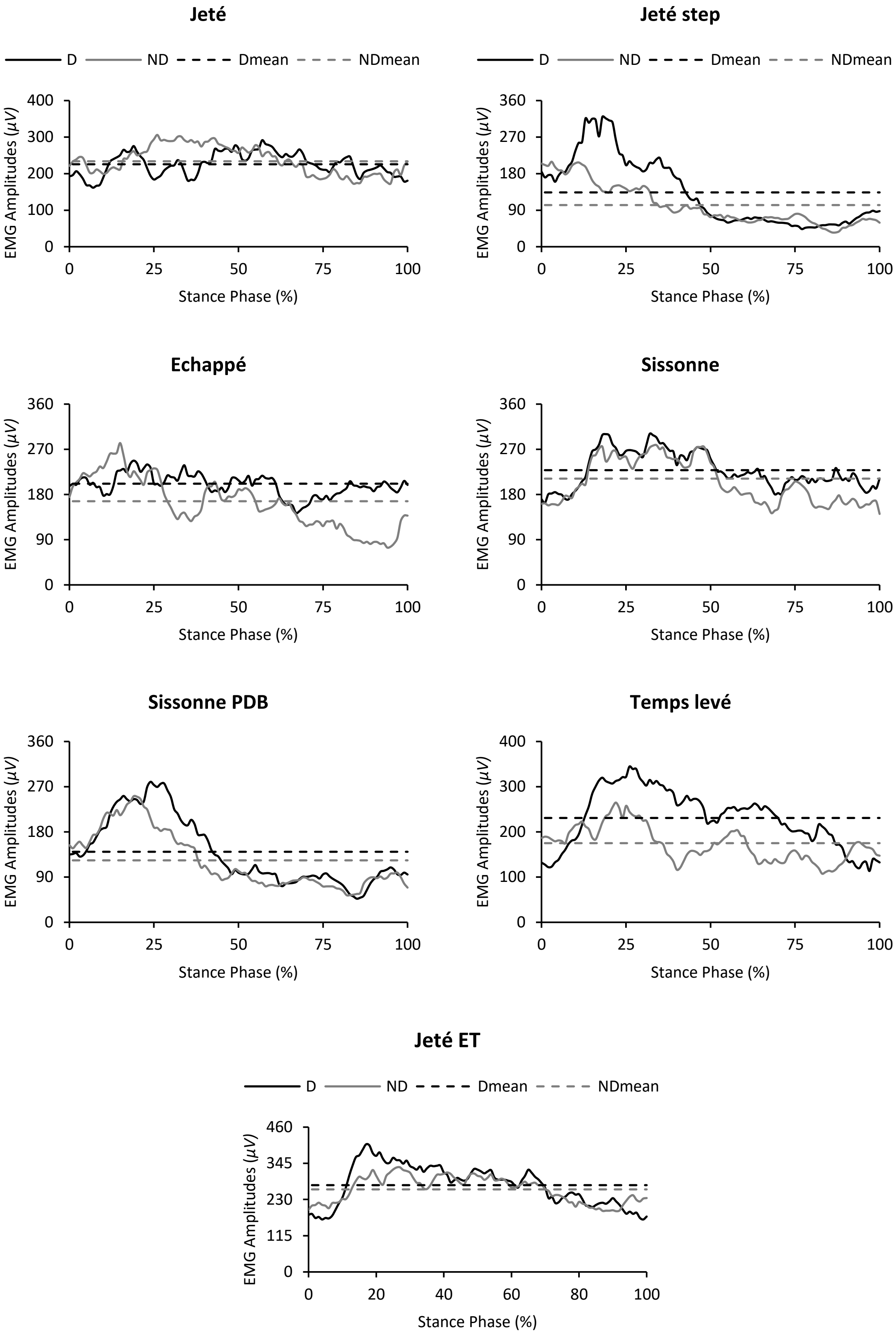

Figure 5.3. Tibialis anterior RMS EMG amplitudes over duration of stance for both the dominant (black solid line) and non-dominant (grey solid line) limb, representing data for all participants and trials. Mean EMG data is presented as dashed lines; dominant (black), non-dominant (grey) limb.

5.4 Discussion

The aim of the current investigation was to quantify the lower limb neuromuscular responses to a battery of ballet-specific movements, specifically, the same as those completed during study 1. A consideration of EMG strengthens the level of analysis, providing a greater insight on the biomechanics of ballet movement, and the muscular strategy used to attenuate GRFs, and control ankle joint kinematics. The kinetic and kinematic observations of study 1 profiled the functional kinesiology requirements of the ankle joint, and therefore supported the selection of the peroneus longus, tibialis anterior and the lateral gastrocnemius muscles in this study.

The first key observation was the significant main effect identified for movement on resultant neuromuscular activity for all muscles. As anticipated, the variation in task demand with regards to planar characteristics, and the hold or transitional landing component had a significant influence on the resultant neuromuscular strategy. Existing literature is not available to substantiate these findings, and thus, generalising the findings beyond the specific methodological design should be treated with caution. The evident hierarchical EMG responses correspond with the kinetic and kinematic data obtained during study 1. For example, the échappé generated the lowest peak $_V$GRFs, whilst variants of the jeté had the highest force magnitudes which has also been shown in previous investigations (Peng et al., 2015; Jarvis and Kulig, 2015). In addition, relative range of motion in the sagittal and coronal plane was smaller during the échappé compared with the jeté-inclusive tasks thereby reflecting lower gross joint displacement. The contrasting mechanics between tasks of the movement battery influences the magnitude of external forces and joint kinematics (Grabowski and Kram, 2008), and thus, each task dictates a bespoke EMG strategy to mediate force. The higher EMG amplitudes in the jump landing tasks with greater kinetic demand, is characteristic of the classic force-EMG relationship (Roberts and Gabaldon, 2008).

Variants of the jeté and sissonne included a transitional element to advance existing experimental protocols comprising a drop and hold task (Orishimo et al., 2009), and better represent true ballet motion. A significant difference between jeté conditions (jeté vs step) was observed for mean EMG, with the transitional-inclusive technique eliciting a 23% reduction in muscle activation. Though not significant, a similar trend towards lower EMG response in the sissonne PDB (9%) compared with the hold variation was also highlighted.

During a land and hold task, co-activation of the ankle musculature is required to stabilise joint kinematics and effectively moderate resultant GRFs (Fox et al., 2008). As demonstrated in the kinematic findings of study 1, the greater overall joint eversion evident in the land and hold tasks may require an increased neuromuscular response to maintain a position of maximal stability (Bonnel et al., 2010). Hence, the lower EMG responses in the transition tasks may be explained by the decreased responsibility of the ankle to provide full support to body weight stabilisation, and rather, to facilitate joint kinematics towards the next movement of a sequence. With an absence of information on EMG responses to a multitude of ballet-specific tasks with varying technical and landing characteristics, further research adopting a similar experimental design is advocated.

Although ballet aesthetics demand a plantarflexed foot configuration during the execution of most techniques (O'Loughlin, Hodgkins and Kennedy, 2008), the instant of ground contact initiates rapid dorsiflexion to manage GRFs as shown in the kinematic findings of study 1. The significantly higher LG response during the jeté step and sissonne PDB compared with the other tasks, is indicative of the transition from amortisation to propulsion with a resultant plantarflexed alignment observed during the terminal stance phase of gait (Stamm and Chiu, 2016). The kinematic observations from study 1 also revealed a progression towards an everted ankle position during the stance phase of all movements. The peroneal muscle EMG response has an important role in preventing the inverted mechanism common to foot and ankle injuries in ballet (Herb et al., 2018). The lowest peak EMG values were observed during the échappé which plausibly reflects the lowest ankle eversion angles highlighted during this movement. In contrast, the highest peak EMG for PL was evident during the jeté to regulate higher GRFs via an everted foot configuration. As the prime regulator of ankle dorsiflexion, TA activation is pivotal towards force absorption during landings (Maharaj, Cresswell, and Lichtwark, 2019). Muscular responses in TA differed significantly across the movement battery, with the greatest EMG activity demonstrated during the jeté ET perhaps given the increased external loads associated with this movement. Ankle dorsiflexion during landing serves as a protective mechanism against injury via decreased joint stiffness, with greater ankle joint excursions vital in reducing GRF magnitudes and lower extremity loading rates (Mason-Mackay, Whatman, and Reid, 2017). The biomechanical data from study 1, and the EMG responses from the current investigation suggest that co-activation of PL and TA may be important towards adopting a pliable landing technique via ankle dorsiflexion and eversion to control resultant GRFs.

A second key observation of the current study was the evident symmetry in EMG response across the battery of ballet-specific movements. That no significant main effects and interactions were identified when the limb factor was involved, demonstrates that symmetry in bilateral muscle activation was independent of task variation. The findings from a previous study (Edwards et al., 2012) similarly support an absence of laterality on resultant EMG parameters during a jump landing manoeuvre. The asymmetric movement profile inherent to ballet emphasises the importance of a balance in limb kinematics between the lower limbs towards performance aesthetics (Kimmerle, 2010). Moreover, injury risk associated with contrasting bilateral muscle activation strategies is lowered (Zifchock et al., 2008). The lack of asymmetry in EMG response between the dominant and non-dominant limbs reflects the kinetic and kinematic findings of study 1. The symmetry in EMG response in this population suggests that similar motor control strategies are adopted to govern the landing component of a jump landing task, regardless of its planar demand. Thus, ballet professionals may infer that movement economy towards performance is the same, irrespective of the limb used to execute the techniques of a routine. This may be the result of training adaptations to ballet classes prescribing alternate limb use when executing the technical components of ballet (Golomer & Fery, 2001). These beneficial developments in bilateral lower limb motor control may have clinical implications towards a reduced injury risk (Bronner & Ojofeitimi, 2006).

Mean and peak EMG amplitudes were quantified in the current study, representing a measure of the average and maximal levels of activity elicited by the muscles during the movement battery (Hibbs et al., 2011). Given the importance of neuromuscular control towards joint kinematics, mean and peak EMG provides insight into the mechanical strategy used to facilitate movement and joint stability during the landing phase. Other common parameters such as mean and median EMG frequencies relate to the spectral-domain of the curve. These frequency measures are typically used to assess muscle fatigue, given that a downward shift on the frequency spectrum would indicate lower muscle unit recruitment (Oskoei and Hu, 2008; Phinyomark et al., 2012). Hence, consideration of the frequency domain may be more suited to investigations on the effects of prolonged exposure to exercise with implications for injury risk, or, to examine muscular recruitment adaptations during rehabilitation. Given that the current study analysed EMG responses to discrete movements in a non-fatigued state, frequency parameters were not presented. However, a secondary analysis of the data revealed no significant differences in median or mean EMG frequency between any of the ballet-specific movements, nor a significant difference between limbs for any task.

It is common practice for EMG data to be normalised, a process which enables the raw amplitudes to be rescaled from microvolts to a percentage, and is achieved by dividing the EMG signal during a particular task by a reference value obtained from the same muscle (Sinclair et al., 2015). The most common reference value used for normalisation is obtained from a Maximal Voluntary Isometric Contraction (MVIC), which should express the maximal neural capacity of a muscle. However, as movement occurs through isotonic muscle contractions, normalised EMG values can often exceed 100% MVIC particularly during rapid, forceful contractions (Burden, 2010). Consequently, this normalisation method is flawed and the value potentially redundant during isotonic contractions. To mitigate against EMG levels above maximal neural capacity (100%), some studies have used the peak EMG value recorded during a task performed at maximum effort. However, maximal force production doesn't necessarily equate to maximal EMG activity. Moreover, different individuals may adopt contrasting neuromuscular control strategies to perform a task, and thus, muscle activity levels cannot be validly compared between muscles or tasks (Halaki and Ginn, 2012). For these reasons, EMG data normalisation was not performed in the current investigation.

There is currently a lack of information available on the EMG responses to ballet movement, and thus, observations from the current study cannot be generalised beyond the specific experimental cohort used. Nevertheless, consideration of neuromuscular activity augments the kinetic and kinematic analyses conducted during study 1, providing insight into the muscular strategy used in joint kinematics when performing ballet manoeuvres. Biomechanics is a multi-modal discipline characterised by an array of assessment tools. Investigating the rigours of performance highlights the biomechanical proficiency required in neuromuscular control, but also in other factors such as strength. A multi-factorial consideration strengthens the level of biomechanics application when profiling the demands of performance, the capacity of dancers, and markers for injury occurrence. Further research incorporating a functionally-relevant testing battery and adopting a multi-modal approach is needed in ballet.

5.5 Conclusion

The current investigation represents the first to quantify the lower limb EMG responses to a range of fundamental ballet-specific jump landing movements in female dancers. The battery of selected tasks was sufficiently sensitive to elicit between-movement differences for all musculature, with EMG responses reflecting the variation in planar demands, and the kinetic and kinematic hierarchy demonstrated in study 1. The magnitudes of muscular activity were proportionate to the mechanics specific to each task, as shown during the échappé and jeté variants. The transitional element of the jeté step and sissonne PBD decreased neuromuscular responses as joint responsibility shifts from full body weight stabilisation on landing, to facilitating joint kinematics towards subsequent movement. The significantly greater PL activation during these tasks reflects functional kinesiology in the presence of plantarflexion at the end of the stance phase. There was no significant bilateral difference in ankle muscle activity, thereby supporting the kinetic and kinematic observations of study 1. The evident symmetry potentially owes to early onset of ballet training and emphasis on bilateral motor coordination. Future investigations in ballet may consider other biomechanical factors such as strength during functionally-relevant movements, with subsequent data potentially informing strength and conditioning practices, whilst providing clinically important information towards injury risk.

Chapter 6. Study three: Isokinetic Strength Profiling of Amateur Female Ballet Dancers

6.1 Introduction

The preceding chapters have quantified the internal (electromyography) and external (ground reaction force) kinetic and kinematic responses to a battery of ballet-specific tasks. The kinematic profile of ballet movement highlighted in study 1, characterised the landing phase of all movements as plantarflexion at initial contact, followed by a combination of dorsiflexion and eversion/inversion to stabilise the ankle during the land and hold techniques, or, to facilitate mobility during the transitional tasks. This movement description is typical of the principal mechanism for ankle trauma (Skazalski et al., 2017). Consistent with the kinematic description of movement technique, the electromyographic (EMG) responses reflected the technical variations and planar demand specific to each task. The neuromuscular demands of ballet performance provides an in-vivo investigation of strength, a routinely purported intrinsic risk factor for ankle injury (Baumhauer et al., 1995; Murphy, Connolly and Beynnon, 2003; Willems et al., 2005).

Strength is a key contributor to force generation during ballet performance, enabling the dancer to execute the many explosive techniques comprising a routine (Watson et al., 2017). In addition, joint strength also plays in important role in managing the ground reaction forces (GRFs) produced during movement. The strength of the peroneal musculature in particular is integral to stabilising the ankle complex by controlling the coronal plane kinematics associated with performing ballet-specific tasks, as demonstrated in study 1 (Fox et al., 2008). The plantarflexed configuration at initial contact undoubtedly manifests injury risk due to the open-pack position and reduced joint stability (Bonnell et al., 2010). However, kinematic analyses have shown ankle trauma to occur in a neutral alignment (Fong et al., 2009) and in dorsiflexion (Kristianslund et al., 2011). Hence, eversion and inversion presents as the primary locomotor contributor to the ankle injury mechanism, and, the risk of injury to dancers is increased in consideration of the coronal plane joint kinematics of ballet movement as highlighted in study 1.

The clinical gold standard method of evaluating joint torque capacity involves isokinetic dynamometry (Dvir and Muller, 2020). Previous assessments of isokinetic ankle strength in non-ballet populations have utilised angular velocities of $30°\cdot s^{-1}$ and $120°\cdot s^{-1}$ (Willems et al., 2002; Pontaga, 2004), representative of slow and fast motions. However, these speeds appear to be selected arbitrarily, without a functional evidence base developed by way of rationale. Moreover, manufacturer guidelines advise a neutral alignment of the foot attachment in the sagittal plane, but the kinaesthetics of ballet performance observes a plantarflexed position which should be considered in some degree during methodological design. Isokinetic data collection is further defined by a predetermined selection of contraction mode, range of motion and angular velocity, which should reflect the specific research question. Study 1 quantified ankle joint angular displacement and velocity during a range of ballet movements, thereby enabling the development of a bespoke isokinetic strength testing protocol to the ballet movement profile, and, the dancer cohort used. The kinematic and kinetic responses also highlighted the technical characteristics of movement, often neglected in isokinetic strength assessment by restricting analyses to a single maximum value defined as peak torque. (Koutedakis and Sharp, 2004; Lim et al., 2015; Tsanaka, Manou and Kellis, 2017). Contemporary research has advocated the inclusion of additional metrics such as the angle at which peak torque is achieved (Small et al., 2010), the angular range over which torque can be maintained (Eustace, Page and Greig, 2017), angle-specific derivatives of strength, and the implications to dynamic strength between contraction modes (El-Ashker et al., 2017). Arguably, strength deficits and angular consideration of joint strength are of greater value for interventions targeting injury reduction and/or performance enhancement. Thus, inclusion of a range of torque metrics assessed over a range of functionally relevant angular displacements and velocities, may provide a more comprehensive understanding of strength capacity of dancers towards ballet movement (Evangelidis, Pain and Folland, 2015; Eustace, Page and Greig, 2017).

In ballet, isokinetic torque evaluations have predominantly focused on the musculature responsible for knee joint function (Westblad, Tsai-Fellander and Johansson, 1995; Koutedakis, 1997; Koutedakis and Sharp, 2004; Kenne and Unnithan, 2008; Lim et al., 2015; Tsanaka, Manou and Kellis, 2017; Lima et al., 2018), with limited information available at the ankle. Methodological designs involving ankle strength assessment in ballet dancers have been restricted to plantar/dorsiflexion protocols (Thomas and Parcel, 2004; Schmitt, Kuni, and Sabo, 2005; Kenne and Unnithan, 2008), and thus, have negated the eversion/inversion displacement common to the injury mechanism, and the movement profile of ballet as shown in study 1. The aim of the current study, therefore, was to comprehensively evaluate bilateral ankle evertor and invertor torque in female ballet dancers across a range of functionally relevant joint angular displacements and velocities. The isokinetic protocol is supported by the sagittal and coronal place kinematic data obtained during study 1. Muscle activity is a key underpinning of torque production, and the magnitude of EMG response has demonstrated a linear relationship with joint strength (Roberts and Gabaldon, 2008). Therefore, a secondary aim was to quantify the resultant neuromuscular responses to the isokinetic strength protocol. Inclusion of EMG analysis enables a comparison of responses between ballet movement and maximal tasks. A thorough profile of ankle eversion and inversion strength that considers bilateral asymmetry in addition to ipsilateral mode and speed-specific asymmetries, will inform clinical interpretation of the training needs required in this cohort.

Experimental research question:

1) Are torque asymmetries evident in female ballet dancer ankle strength during a functionally relevant testing protocol following identification of a dominant limb?

6.2 Methods

Participants

Although the participants for this study were the same female ballet cohort used in studies 1 and 2, a power calculation (G*Power v.3.1, Heinrich-Heine-Universität, Dusseldorf, Germany) was conducted to qualify the power of the sample, with measures in concentric eversion/inversion peak torque and functional range used. The power analysis revealed that the cohort of 14 was sufficient to elicit an observed statistical power of 0.8 for significant differences ($p < 0.05$) between the dominant and non-dominant limb. All participants met the inclusion criteria, and completed pre-screening health assessments in accordance with the protocols outlined in Chapter 3.

Procedures

All participants were required to attend the Musculoskeletal Laboratory for one experimental testing session. Participants initially completed the standardised warm-up procedure described in Chapter 3, and followed this by performing 10 slow eversion and inversion repetitions (seated with legs outstretched) for both limbs. Participants then completed five sub-maximal (self-determined 50% effort) trials of concentric ankle eversion (CON_{EV}) and inversion (CON_{INV}), and, eccentric ankle inversion (ECC_{INV}) at all experimental testing velocities as part of the warm-up protocol. The familiarisation trials were completed with progressive increments in angular velocity through the sequence; 30°·s^{-1}, 60°·s^{-1}, 90°·s^{-1}, and 120°·s^{-1} for both limbs. Experimental trials were subsequently completed following a five-minute rest period, with five maximal repetitions for each contraction mode and speed.

Isokinetic strength assessment

An isokinetic dynamometer (System 4 pro, Biodex Medical Systems, Shirley, New York, USA) and its respective manufacturer software programme was used to assess bilateral ankle eversion and inversion strength, and, determine various torque-related outcome measures. To optimise data validity and reliability, calibration of the isokinetic dynamometer was conducted before the start of each experimental testing session. Limb dominance for each participant was determined using the same method described in Chapter 4.2. Isokinetic assessments of ankle inversion and eversion strength using the Biodex dynamometer have been demonstrated to be highly reliable (Intratester ICCs for inversion = 0.92-0.96; for eversion = 0.87-0.9) (Aydog et al., 2004). All participants were secured in a seated position, and the isokinetic dynamometer was configured according to manufacturer's guidelines for ankle eversion/inversion strength assessment (seat orientation, 90°; seatback tilt, 70°; dynamometer head orientation, 0°; dynamometer head tilt, 50°). Each participants' foot was fixed to the relevant dynamometer attachment using the Velcro straps provided. A goniometer was used to set the foot attachment in 20° of plantarflexion to partially replicate the orientation of the of foot when landing from a ballet-specific jump landing tasks, as shown in the sagittal plane kinematic data of study 1. To isolate the ankle joint and prevent extraneous segmental motion from influencing torque production, dynamometer straps were applied across the chest and the anterior mid-thigh of the contralateral limb. The attachment typically placed under the gastrocnemius was modified, and positioned at the mid-portion of the posterior thigh to secure the upper leg. This configuration enabled EMG analysis of the ankle evertor/invertor musculature to be conducted during the trials. To exercise a standardised test protocol, ankle eversion and inversion motion limits were set at 20° from a neutral position (vertical alignment of the foot in the coronal plane), resulting in an overall range of motion of 40°.

Initiated at a position of max inversion (20°), all participants completed five maximal concentric ankle eversion and inversion trials (Sekir et al., 2007), at angular velocities of

$90°·s^{-1}$, $60°·s^{-1}$, $120°·s^{-1}$ and $30°·s^{-1}$ for both limbs, in accordance with previous recommendations (Fish, Milligan and Killey, 2014). The non-linear order was chosen to minimise any potential learning effect. Pilot testing revealed that no isokinetic phase was determined at angular velocities $\geq 150°·s^{-1}$, and therefore, $120°·s^{-1}$ was selected as the fastest speed with an extended isokinetic period. Hence, the greatest isokinetic speed used in the current study ($120°·s^{-1}$) was appropriate. The same procedure was then completed for the eccentric ankle inversion trials. Concentric ankle eversion and inversion trials at each angular velocity were interspersed with a one-minute rest period, whilst 10-minutes rest separated ipsilateral concentric and eccentric trials to minimise the accumulation of fatigue (Yuksel et al., 2011). No performance feedback was presented during any of the experimental trials. The selected overall joint displacement and angular velocities of the isokinetic strength protocol were supported by coronal plane kinematic information obtained during study 1, which is presented in Table 6.1.

Table 6.1. Bilateral coronal plane kinematic data to support the isokinetic strength testing protocol.

Variable	Limb	Movement						
		Jeté	**Jeté Step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
Total ROM (°)*	Dom	24	19	16	25	28	20	25
	Non-Dom	23	17	13	24	27	18	24
Time at ≤ 120°·s-1	Dom	78	51	87	82	65	77	89
	Non-Dom	80	50	87	82	69	77	86

Dom, Dominant; Non-Dom, Non-dominant; ROM, Range of motion * to the nearest whole degree; $°.s^{-1}$, Degrees per second.

EMG analysis

Lower limb muscle activity during the isokinetic strength protocol was quantified using an EMG system (Noraxon, Noraxon USA inc, Arizona, USA). The protocol for muscle selection, skin preparation and electrode placement was in accordance with the methods outlined in Chapter 5. Corresponding signals were transmitted telemetrically to a nearby desktop DTS receiver, and accompanying EMG system software (MyoResearch XP, Noraxon USA inc, Arizona, USA) was used to record, store and process resultant EMG responses.

Data Processing

Raw torque-angle time history data from each limb, contraction mode and angular velocity were exported to Excel (Microsoft Corporation, Washington, USA) for further analysis. With torque overshoot removed, the isokinetic phase of each repetition was determined, and the repetition producing the highest torque was analysed. At each velocity and mode of contraction, Peak Torque (PT), corresponding Angle of Peak Torque (APT), and Functional Range (FR - defined as the range over which 85% of peak torque is maintained, Croisier et al., 2008) was established. Angle-Specific Torque (AST) data were calculated in 5° increments across the entire angular range (40°) for all angular velocities and contraction modes. Dynamic Control Ratios (DCRs) between the contraction modes were calculated (i.e. CON_{EV}:CON_{INV}), and defined using PT (DCR_{PT}) and AST (DCR_{AST}) values.

The neuromuscular responses corresponding to the selected isokinetic repetition were analysed in the relevant software (MyoResearch XP, Noraxon USA inc, Arizona, USA). The EMG signals were initially bandpass filtered using 20 Hz (high-pass) and 450 Hz (low-pass) cut-offs to remove alias data in the form of movement artefact and high frequency noise. Data was full-wave rectified and smoothed using the Root Mean Squared (RMS) method of a 75-millisecond time constant. Resultant EMG envelopes were then analysed to obtain mean and peak EMG amplitudes in accordance with study 2, and integrated EMG (iEMG) which represents the area under the EMG curve.

Statistical Analysis

Descriptive statistics are presented as mean ± σ. The distribution of data was quantified using histograms, Q-Q plots, skewness and kurtosis, and the Shapiro-Wilk statistic. With the data normality assumption satisfied, linear mixed models were employed to examine bilateral isokinetic strength differences in each outcome measure across all testing velocities and contraction modes. Linear mixed models were chosen to account for missing data at selected angular displacements and velocities (Peng and Lu, 2012). On the contrary, general linear models deal with missing data points through listwise deletion, and thus, one missing data point across the selected variables would have resulted in the removal of all data relating to the specific participant. Repeated measures ANOVAs were deployed to investigate main effects and interactions for the selected EMG metrics. Bonferroni-corrected post-hoc pairwise comparisons for significant main effects and interactions were determined as required, and 95% CIs and Cohen's *d* effect sizes (small, 0.20-0.49; moderate, 0.50-0.79; large > 0.80) are also presented (Cohen, 1988). Alpha was determined *a priori* and deemed statistically significant at the $p < 0.05$ level for all outcome measures. Statistical analyses were conducted using IBM SPSS statistics V25.0 software (IBM, Armonk, New York, USA).

6.3 Results

Isokinetic strength

Peak torque

Figure 6.1 summarises the influence of contraction mode and angular velocity on bilateral PT. Significant main effects were identified for contraction mode ($F = 71.91$, $p < 0.01$) and angular velocity ($F= 8.00$, $p < 0.01$), and the contraction mode *x* angular velocity interaction was also significant ($F = 4.90$, $p < 0.01$). Figure 6.1 demonstrates that ECC_{INV} PT was significantly greater than CON_{EV} and CON_{INV} at 60°·s^{-1} ($p < 0.01$, $d = 0.43$-0.48), 90°·s^{-1} ($p < 0.01$, $d = 0.67$-0.73) and 120°·s^{-1} ($p < 0.01$, $d = 0.73$-0.77). Analyses revealed no significant difference between CON_{EV} and CON_{INV} PT at all testing velocities ($p = 1.00$) There was no significant main effect for limb ($F = 0.89$, $p = 0.35$), nor any significant limb *x* contraction mode ($F = 0.33$, $p = 0.72$), limb *x* angular velocity ($F = 0.10$, $p = 0.96$), or limb *x* contraction mode *x* angular velocity ($F = 0.11$, $p = 1.00$) interaction. In the subsequent metrics, for instances where the significant main effects/interactions involving limb are not significant, corresponding values for significant contraction mode, angular velocity and angle main effects/interactions represent an average from the dominant and non-dominant limb.

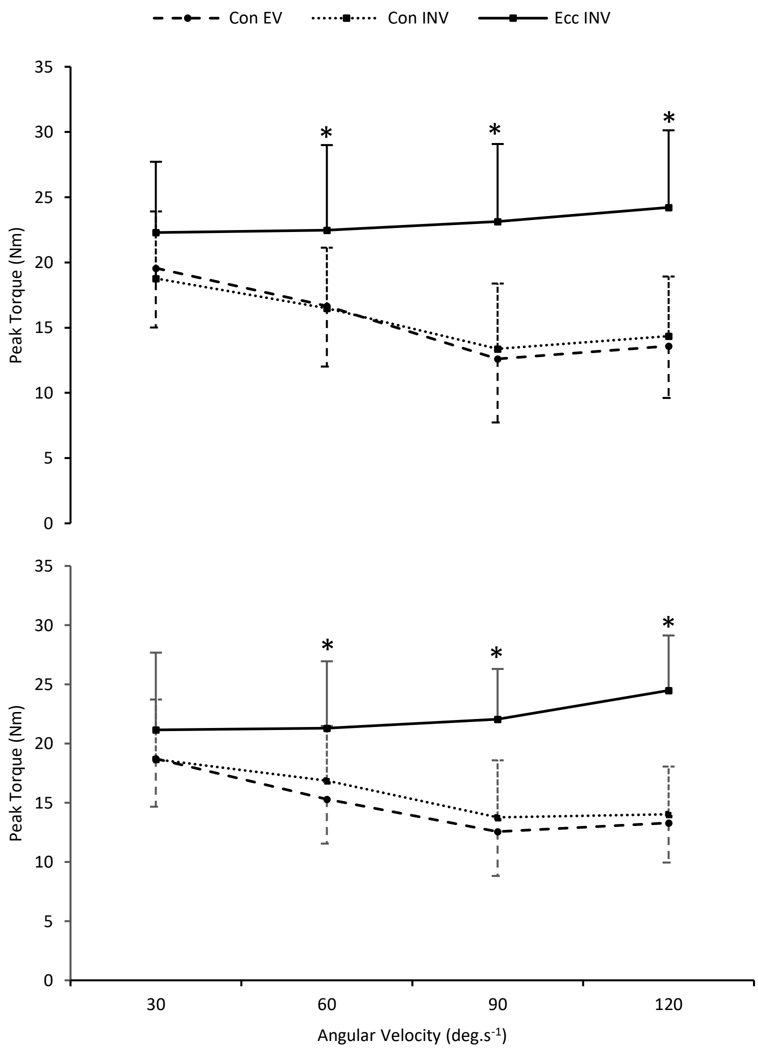

Figure 6.1. PT for each mode of contraction for the dominant (top graph) and non-dominant (bottom graph) limb. Values are mean ± σ. * denotes a significant difference between the eccentric and concentric-inclusive contraction modes.

Angle of peak torque

Table 6.2 displays bilateral APT data for all contraction modes and angular velocities. Analyses revealed a significant main effect for contraction mode ($F = 71.91$, $p < 0.01$), with ECC_{INV} APT (27.1 ± 7.2°; CI: 25.8-28.5°) occurring significantly later in the range of motion compared with CON_{EV} (18.5 ± 6.1°; CI: 17.2-19.7°, $p < 0.01$, $d = 0.55$) and CON_{INV} (16.6 ± 6.5°; CI: 15.4-17.9°, $p < 0.01$, $d = 0.61$) irrespective of angular velocity. Further, a significant main effect for angular velocity ($F = 3.51$, $p = 0.02$) demonstrated that APT was achieved significantly earlier at $30°\cdot s^{-1}$ (18.8 ± 9.0°, CI: 17.4-20.3°) compared with $90°\cdot s^{-1}$ (22.2 ± 7.6°; CI: 20.7-23.6°, $p = 0.01$, $d = 0.19$) irrespective of contraction type. No significant contraction mode *x* angular velocity ($F = 1.28$, $p = 0.27$) interaction was observed. The linear mixed model highlighted no significant main effect for limb ($F = 0.05$, $p = 0.82$), and there was no significant limb *x* contraction mode (F = 0.76, p = 0.46), limb *x* angular velocity (F = 0.7, p = 0.55), or limb *x* contraction mode *x* angular velocity (F = 0.39, p = 0.89) interaction.

Table 6.2. The influence of angular velocity, limb and mode of contraction on APT. Corresponding values are mean ± σ.

Angular Velocity	Limb	Contraction Mode		
		CON_{EV} APT (°)	CON_{INV} APT (°)	ECC_{INV} APT (°)*
$30°\cdot s^{-1}$	Dominant	15.8 ± 6.6	14.4 ± 7.1	27.3 ± 9.8
	Non-Dominant	13.9 ± 5.2	15.5 ± 7.0	26.0 ± 8.3
$60°\cdot s^{-1}$	Dominant	20.2 ± 6.6	17.8 ± 5.4	26.5 ± 7.1
	Non-Dominant	19.1 ± 6.7	16.9 ± 5.8	24.5 ± 6.5
$90°\cdot s^{-1\delta}$	Dominant	20.6 ± 8.0	16.3 ± 6.3	27.1 ± 6.8
	Non-Dominant	20.9 ± 6.1	20.4 ± 5.7	27.1 ± 6.4
$120°\cdot s^{-1}$	Dominant	17.1 ± 7.0	15.2 ± 7.3	29.6 ± 6.7
	Non-Dominant	20.0 ± 5.0	16.5 ± 6.7	27.6 ± 5.6

CON_{EV}, Concentric Eversion; CON_{INV}, Concentric Inversion; ECC_{INV}, Eccentric Inversion; APT, Angle of Peak Torque; °, degrees; $°\cdot s^{-1}$, degrees per second. * denotes a significant difference with the corresponding CON_{EV} and CON_{INV} contractions irrespective of angular velocity. δ denotes a significant difference between $30°\cdot s^{-1}$ and $90°\cdot s^{-1}$ irrespective of contraction mode.

Functional range

Figure 6.2 illustrates the influence of contraction mode and angular velocity on bilateral FR. No significant main effect for contraction mode was found (F = 1.89, p = 0.15), however a significant main effect for angular velocity (F = 17.37, p < 0.01) was revealed irrespective of contraction mode. Post-hoc analyses showed that FR at $30°\cdot s^{-1}$ (18.57 ± 5.95°; CI: 17.6-19.8°) was significantly lower than at $60°\cdot s^{-1}$ (21.1 ± 5.0°; CI: 20.0-22.2°; p = 0.02, *d* = 0.2) but significantly higher compared with $120°\cdot s^{-1}$ (15.8 ± 6.3°; CI: 14.2-16.5°; p < 0.01, *d* = 0.22). FR at $60°\cdot s^{-1}$ was significantly greater than at $120°\cdot s^{-1}$ (p < 0.01, *d* = 0.41), and at $90°\cdot s^{-1}$ (19.4 ± 5.4°; CI: 18.2-20.4°) compared with $120°\cdot s^{-1}$ (p < 0.01, *d* = 0.29). The significant contraction mode *x* angular velocity interaction (p < 0.01) demonstrated that ECC_{INV} FR was significantly greater at than CON_{EV} and CON_{INV} at $30°\cdot s^{-1}$ (p < 0.01, *d* = 0.46-0.57), but significantly lower at $120°\cdot s^{-1}$ (p < 0.01, *d* = 0.67-0.73). CON_{EV} and CON_{INV} FR was not significantly different at all testing velocities (P ≥ 0.79). There was no significant main effect for limb (F = 2.69, p = 0.10), and no significant limb *x* contraction mode (F =

0.42, p = 0.66), limb *x* angular velocity (F = 0.02, p = 1.00), or limb *x* contraction mode *x* angular velocity (F = 0.25, p = 0.96) interaction.

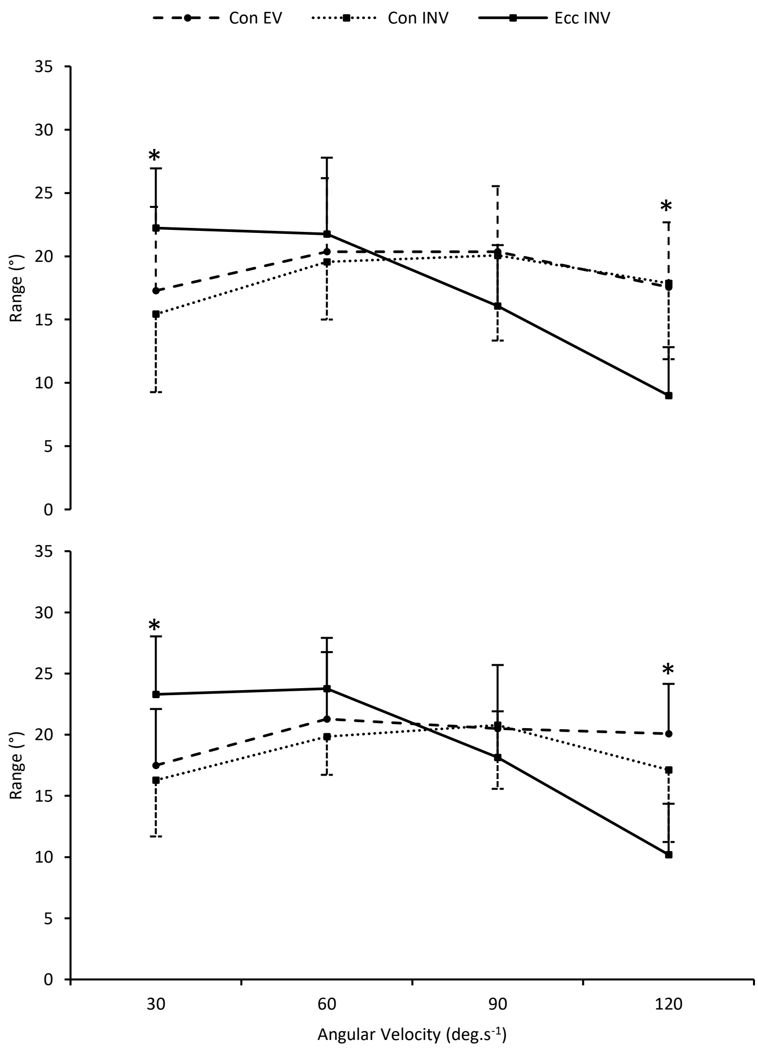

Figure 6.2. FR for each mode of contraction for the dominant (top graph) and non-dominant (bottom graph) limb. Values are mean ± σ. * denotes a significant difference between the eccentric and concentric-inclusive contraction modes.

Dynamic control ratios calculated from PT data.

DCR_{PT} values are presented in Table 6.3. Significant main effects for contraction mode (F = 56.46, $p < 0.01$) and angular velocity (F = 26.09, $p < 0.01$), and the corresponding contraction mode *x* angular velocity interaction (F = 2.80, p = 0.01) were highlighted. There was no indication of any contraction mode *x* angular velocity interactions at $30°\cdot s^{-1}$, however CON_{EV}:CON_{INV} dynamic controls ratios were significantly greater than CON_{EV}:ECC_{INV} and CON_{INV}:ECC_{INV} at $60°\cdot s^{-1}$ (p = 0.01, *d* = 0.38-0.49), $90°\cdot s^{-1}$ ($p < 0.01$, *d* = 0.70-0.77) and

120°·s^{-1} $p < 0.01$, $d = 0.74$-0.78) respectively. The CON_{EV}:ECC_{INV} and the CON_{INV}:ECC_{INV} dynamic control ratios were not significant different at all testing velocities ($p = 1.00$) There was no significant main effect for limb ($F = 0.02$, $p = 0.90$), nor a significant limb *x* contraction mode ($F = 1.90$, $p = 0.15$), limb *x* angular velocity ($F = 0.41$, $p = 0.75$), or limb *x* contraction mode *x* angular velocity ($F = 0.20$, $p = 0.98$) interaction.

Table 6.3. The influence of limb and angular velocity on selected Dynamic Control Ratios calculated from PT. Values are mean ± σ.

Dynamic Control Ratio	Limb	Angular Velocity			
		30°·s^{-1}	60°·s^{-1}	90°·s^{-1}	120°·s^{-1}
CON_{EV}:	Dominant	1.08 ± 0.27	1.05 ± 0.26*	0.96 ± 0.18*	0.97 ± 0.18*
CON_{INV}	Non-Dominant	1.02 ± 0.19	0.92 ± 0.12*	0.96 ± 0.18*	0.96 ± 0.13*
CON_{EV}:	Dominant	0.91 ± 0.26	0.77 ± 0.26	0.55 ± 0.21	0.58 ± 0.17
ECC_{INV}	Non-Dominant	0.94 ± 0.27	0.74 ± 0.12	0.57 ± 0.12	0.59 ± 0.15
CON_{INV}:	Dominant	0.86 ± 0.27	0.78 ± 0.30	0.58 ± 0.20	0.61 ± 0.16
ECC_{INV}	Non-Dominant	0.98 ± 0.37	0.82 ± 0.16	0.63 ± 0.15	0.63 ± 0.20

CON_{EV}, Concentric Eversion; CON_{INV}, Concentric Inversion; ECC_{INV}, Eccentric Inversion; °·s^{-1}, degrees per second. D, Dominant; ND, Non-Dominant. * denotes a significant difference with the corresponding ECC_{INV} – inclusive ratios.

Angle-specific torque

Figure 6.3 depicts the influence of contraction mode, angular velocity, and angle on bilateral AST. Significant main effects for contraction mode ($F = 212.82$, $p < 0.01$), angle ($F = 10.39$, $p < 0.01$), and angular velocity ($F = 15.59$, $p < 0.01$), and, significant contraction mode *x* angle ($F = 2.91$, $p < 0.01$), and contraction mode *x* angular velocity ($F = 14.52$, $p < 0.01$) interactions were identified. Analyses revealed that ECC_{INV} torque was significantly greater than the two concentric modes for angles ≥ 15° during the 30°·s^{-1}, 60°·s^{-1} and 90°·s^{-1} trials, and for angles ≥ 25° during 120°·s^{-1} (see Figure 6.3). There was no significant main effect for limb ($F = 0.29$, $p = 0.59$), nor a significant interaction for limb *x* contraction mode ($F = 0.16$, $p = 0.86$), limb *x* angle ($F = 0.09$, $p = 1.00$), limb *x* angular velocity ($F = 0.11$, $p = 0.95$), limb *x* contraction mode *x* angle ($F = 0.21$, $p = 1.00$), limb *x* contraction mode *x* angular velocity ($F = 0.66$, $p = 0.68$), limb *x* angle *x* angular velocity ($F = 0.06$, $p = 1.00$), or limb *x* contraction mode *x* angle *x* angular velocity ($F = 0.07$, $p = 1.00$).

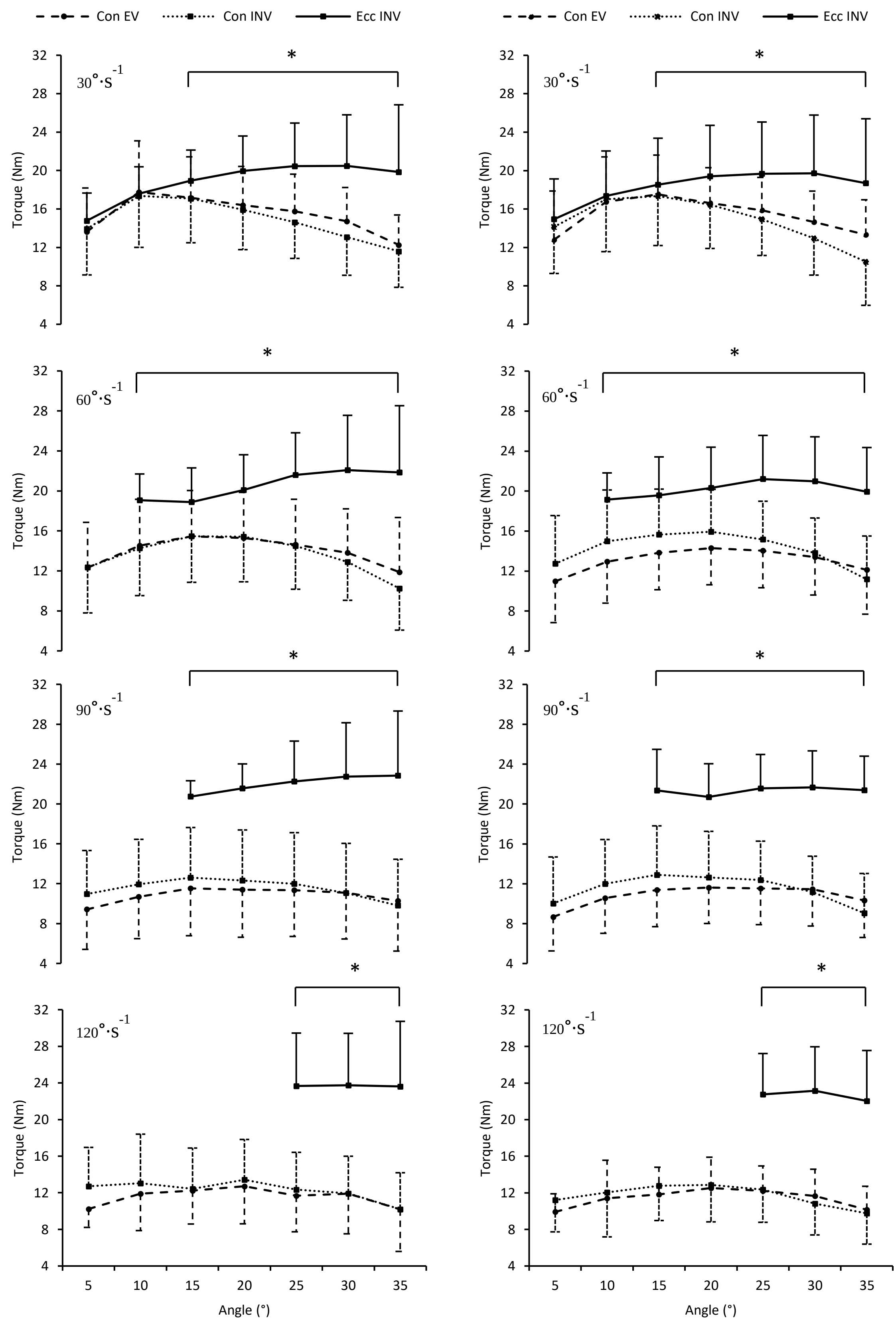

Figure 6.3. Angle-specific torque for each mode of contraction for the dominant (left graphs) and non-dominant (right graphs) limb. Values are mean ± σ. * denotes a significant difference for AST between the eccentric and concentric-inclusive contraction modes.

Dynamic control ratios derived from AST data.

Table 6.4 summarises the effect of contraction mode, angular velocity and angle on the respective bilateral DCR_{AST}. Significant main effects for contraction mode ($F = 183.95$, $p < 0.01$), angle ($F = 3.04$, $p = 0.01$) and angular velocity ($F = 34.48$, $p < 0.01$) were identified, along with a significant contraction mode *x* angle ($F = 11.81$, $p < 0.01$) and contraction mode *x* angular velocity ($F = 2.63$, $p = 0.02$) interaction. At angles ≥ 15°, CON_{EV}:CON_{INV} dynamic control ratios were significantly higher than CON_{EV}:ECC_{INV} and CON_{INV}:ECC_{INV}. Moreover, CON_{EV}:CON_{INV} dynamic control ratios were also significantly greater than the ECC_{INV}-inclusive ratios at all isokinetic speeds (see Table 6.3). There was no significant angle *x* angular velocity ($F = 0.89$, $p = 0.59$) or contraction mode *x* angle *x* angular velocity ($F = 0.52$, $p = 0.97$) interaction. There was no significant main effect for limb ($F = 0.31$, $p = 0.58$), nor a significant limb *x* contraction mode ($F = 4.25$, $p = 0.08$), limb *x* angle ($F = 0.11$, $p = 1.00$), limb *x* angular velocity ($F = 1.66$, $p = 0.17$), limb *x* contraction mode *x* angle ($F = 0.2$, $p = 1.00$), limb *x* contraction mode *x* angular velocity ($F = 0.94$, $p = 0.47$), limb *x* angle *x* angular velocity ($F = 0.23$, $p = 1.00$), or limb *x* contraction mode *x* angle *x* angular velocity ($F = 0.22.$, $p = 1.00$) interaction.

Table 6.4. The influence of limb, angle and angular velocity on the corresponding DCR$_{AST}$ data. Values are mean ± σ.

	Angular Velocity							
CON$_{EV}$: CON$_{INV}$	**30°·s^{-1}**		**60°·s^{-1}**		**90°·s^{-1}**		**120°·s^{-1}**	
Angle (°)	D	ND	D	ND	D	ND	D	ND
5	1.01 ± 0.32	0.95 ± 0.33	1.08 ± 0.39	0.88 ± 0.22	0.85 ± 0.28	0.80 ± 0.30	0.86 ± 0.22	0.92 ± 0.19
10	1.05 ± 0.26	1.01 ± 0.23	1.08* ± 0.32	0.87* ±0.13	0.92 ± 0.21	0.90 ± 0.20	0.96 ± 0.21	0.97 ± 0.20
15	1.04 ± 0.26	1.04 ± 0.23	1.04* ± 0.28	0.89* ± 0.11	0.95 ± 0.23	0.90 ± 0.18	1.05 ± 0.30	0.94 ± 0.12
20	1.08* ± 0.29	1.04* ± 0.22	1.03* ± 0.27	0.90* ± 0.12	0.97* ± 0.25	0.94* ± 0.16	0.97 ± 0.21	0.99 ± 0.15
25	1.12* ± 0.29	1.08* ± 0.21	1.05* ± 0.30	0.92* ± 0.13	1.01* ± 0.32	0.93* ± 0.14	0.98* ± 0.25	1.01* ± 0.14
30	1.20* ± 0.38	1.16* ± 0.23	1.12* ± 0.36	0.97* ± 0.16	1.09* ± 0.38	1.02* ± 0.20	1.03* ± 0.29	1.14* ± 0.40
35	1.22* ± 0.56	1.30* ± 0.44	1.32* ± 0.67	1.18* ± 0.66	1.17* ± 0.58	1.27* ± 0.66	1.07* ± 0.34	1.06* ± 0.20
CON$_{EV}$: ECC$_{INV}$								
5	0.93 ± 0.31	0.91 ± 0.36						
10	1.03 ± 0.31	1.04 ± 0.33	0.81 ± 0.34	0.71 ± 0.18				
15	0.91 ± 0.18	1.00 ± 0.26	0.83 ± 0.26	0.74 ± 0.15				
20	0.82 ± 0.13	0.89 ± 0.20	0.76 ± 0.21	0.73 ± 0.14	0.54 ± 0.23	0.57 ± 0.14		
25	0.78 ± 0.13	0.84 ± 0.20	0.70 ± 0.18	0.69 ± 0.11	0.53 ± 0.20	0.56 ± 0.15	0.59 ± 0.22	0.55 ± 0.11
30	0.73 ± 0.14	0.65 ± 0.18	0.66 ± 0.19	0.67 ± 0.12	0.51 ± 0.19	0.55 ± 0.14	0.54 ± 0.17	0.55 ± 0.12
35	0.64 ± 0.26	0.77 ± 0.25	0.58 ± 0.26	0.62 ± 0.17	0.50 ± 0.24	0.50 ± 0.12	0.47 ± 0.16	0.50 ± 0.14
CON$_{INV}$: ECC$_{INV}$								
5	0.95 ± 0.31	1.00 ± 0.31						
10	0.99 ± 0.30	1.07 ± 0.37	0.75 ± 0.27	0.84 ± 0.19				
15	0.91 ± 0.27	1.02 ± 0.39	0.84 ± 0.31	0.84 ± 0.16				
20	0.81 ± 0.25	0.93 ± 0.36	0.79 ± 0.30	0.82 ± 0.15	0.56 ± 0.22	0.61 ± 0.16		
25	0.73 ± 0.18	0.83 ± 0.29	0.72 ± 0.26	0.77 ± 0.15	0.55 ± 0.22	0.60 ± 0.14	0.60 ± 0.16	0.58 ± 0.15
30	0.65 ± 0.17	0.72 ± 0.23	0.64 ± 0.21	0.71 ± 0.18	0.51 ± 0.21	0.53 ± 0.13	0.53 ± 0.17	0.50 ± 0.16
35	0.58 ± 0.15	0.63 ± 0.29	0.53 ± 0.25	0.60 ± 0.21	0.47 ± 0.22	0.45 ± 0.18	0.48 ± 0.18	0.50 ± 0.20

CON$_{EV}$, Concentric Eversion; CON$_{INV}$, Concentric Inversion; ECC$_{INV}$, Eccentric Inversion; D, Dominant; ND, Non-Dominant; °, degrees; °·s^{-1}, degrees per second. * denotes a significant difference with the corresponding ECC$_{INV}$ – inclusive rations. Grey-shaded area represents the angular displacements and velocities without an isokinetic phase.

EMG responses

Amplitude Domain

Table 6.5 summarises the influence of laterality and angular velocity on EMG response for each muscle and contraction mode. The ANOVA returned a non-significant main effect for contraction mode on mean (F = 1.03, p = 0.37) and peak EMG (F = 0.26, p = 0.77), but contraction mode was significant for iEMG (F = 33.20, p < 0.01). There was no significant main effect for angular velocity on mean (F = 0.87, p = 0.47) or peak EMG (F = 1.86, p = 0.15), but it was significant for iEMG. (F = 38.93, p < 0.01). The was also a significant main effect identified for muscle on mean (F = 63.71, p < 0.01), peak (F = 57.76, p = 0.001), and iEMG (F = 52.90, p < 0.01). However, there was no significant main effect for limb on mean (F = 3.90, p = 0 .07), peak (F = 2.23, p = 0.16), or iEMG (F = 2.84, p = 0.12).

Table 6.5 Bilateral responses in selected EMG amplitude parameters during the isokinetic testing protocol. Corresponding values are to the nearest whole µV, expressed as mean ± σ.

CON_{EV}					
Muscle *Parameter*	**Limb**	**Angular Velocity**			
LG		**30°·s^{-1}**	**60°·s^{-1}**	**90°·s^{-1}**	**120°·s^{-1}**
Mean*	Dom	20 ± 15	21 ± 17	19 ± 16	18 ± 13
(µV)	Non-Dom	19 ± 12	17 ± 9	18 ± 11	19 ± 11
Peak$^{\delta}$	Dom	43 ± 30	41 ± 41	35 ± 40	37 ± 26
(µV)	Non-Dom	38 ± 25	26 ± 14	35 ± 31	31 ± 18
iEMG	Dom	28 ± 21	17 ± 15	12 ± 10	10 ± 6
*(uV*s)*	Non-Dom	25 ± 17	14 ± 6	11 ± 6	10 ± 6
PL					
Mean	Dom	54 ± 31	57 ± 36	52 ± 33	53 ± 34
(µV)	Non-Dom	77 ± 68	79 ± 64	74 ± 62	74 ± 56
Peak	Dom	96 ± 63	96 ± 72	97 ± 62	91 ± 71
(µV)	Non-Dom	132 ± 117	132 ± 108	120 ± 99	107 ± 75
iEMG	Dom	75 ± 49	46 ± 29	36 ± 27	30 ± 22
*(uV*s)*	Non-Dom	102 ± 96	67 ± 55	51 ± 47	39 ± 30
TA					
Mean	Dom	83 ± 50	88 ± 47	67 ± 44	82 ± 44
(µV)	Non-Dom	84 ± 50	79 ± 43	60 ± 36	80 ± 47
Peak	Dom	147 ± 83	153 ± 78	126 ± 84	135 ± 67
(µV)	Non-Dom	154 ± 88	134 ± 69	106 ± 55	133 ± 72
iEMG	Dom	116 ± 78	72 ± 42	43 ± 29	45 ± 25
*(uV*s)*	Non-Dom	112 ± 76	67 ± 34	40 ± 27	41 ± 23

Table 6.5. *Continued*

CON$_{INV}$					
LG					
Mean*	Dom	24 ± 20	21 ± 19	26 ± 32	26 ± 24
(µV)	Non-Dom	29 ± 24	28 ± 24	27 ± 34	28 ± 25
Peak$^\delta$	Dom	46 ± 36	42 ± 37	46 ± 67	46 ± 44
(µV)	Non-Dom	48 ± 39	48 ± 46	45 ± 61	46 ± 41
iEMG	Dom	34 ± 31	19 ± 19	18 ± 24	15 ± 15
*(uV*s)*	Non-Dom	39 ± 36	25 ± 23	20 ± 30	17 ± 17
PL					
Mean	Dom	60 ± 31	55 ± 28	63 ± 41	64 ± 30
(µV)	Non-Dom	66 ± 31	69 ± 32	73 ± 29	64 ± 26
Peak	Dom	107 ± 54	98 ± 40	111 ± 74	107 ± 48
(µV)	Non-Dom	121 ± 65	126 ± 42	113 ± 42	110 ± 49
iEMG	Dom	83 ± 56	51 ± 30	44 ± 30	35 ± 17
*(uV*s)*	Non-Dom	87 ± 48	63 ± 31	50 ± 20	38 ± 18
TA					
Mean	Dom	88 ± 52	77 ± 58	73 ± 52	84 ± 54
(µV)	Non-Dom	81 ± 46	88 ± 52	79 ± 56	74 ± 45
Peak	Dom	151 ± 71	133 ± 93	127 ± 101	137 ± 82
(µV)	Non-Dom	153 ± 73	156 ± 99	146 ± 102	128 ± 66
iEMG	Dom	122 ± 79	68 ± 52	51 ± 40	46 ± 30
*(uV*s)*	Non-Dom	109 ± 74	77 ± 45	55 ± 40	45 ± 30
ECC$_{INV}$$^\gamma$					
LG					
Mean*	Dom	12 ± 8	12 ± 10	11 ± 9	11 ± 11
(µV)	Non-Dom	12 ± 6	11 ± 8	12 ± 7	12 ± 6
Peak$^\delta$	Dom	23 ± 13	21 ± 16	20 ± 14	18 ± 8
(µV)	Non-Dom	23 ± 10	19 ± 13	26 ± 11	21 ± 9
iEMG	Dom	28 ± 15	21 ± 17	19 ± 16	19 ± 14
*(uV*s)*	Non-Dom	32 ± 10	20 ± 15	23 ± 14	22 ± 14
PL					
Mean	Dom	54 ± 26	55 ± 30	59 ± 31	58 ± 45
(µV)	Non-Dom	76 ± 46	74 ± 38	83 ± 40	82 ± 32
Peak	Dom	102 ± 43	112 ± 47	116 ± 45	98 ± 28
(µV)	Non-Dom	157 ± 83	154 ± 80	170 ± 83	167 ± 76
iEMG	Dom	134 ± 57	104 ± 54	104 ± 56	102 ± 52
*(uV*s)*	Non-Dom	208 ± 126	159 ± 152	189 ± 194	156 ± 74
TA					
Mean	Dom	64 ± 22	59 ± 24	62 ± 19	65 ± 30
(µV)	Non-Dom	69 ± 42	81 ± 64	73 ± 38	68 ± 23
Peak	Dom	128 ± 53	116 ± 44	123 ± 38	128 ± 68
(µV)	Non-Dom	139 ± 86	156 ± 105	137 ± 62	145 ± 53
iEMG	Dom	159 ± 56	116 ± 60	112 ± 37	124 ± 68
*(uV*s)*	Non-Dom	194 ± 127	151 ± 118	148 ± 105	129 ± 56

TA, Tibialis Anterior; PL, Peroneus longus; LG, Lateral gastrocnemius; *µV*, Microvolts; *µV*s*, Microvolts per second; Dom, Dominant; Non-Dom, Non-Dominant * indicates that LG responses is significantly ($p < 0.05$) lower than PL and TA at all angular velocities. $^\delta$ indicates that LG responses is significantly ($p < 0.05$) lower than PL and TA at all angular velocities. $^\gamma$ signifies that EMG response in the non-dominant limb wee significant ($p < 0.05$) higher than the dominant leg, irrespective of angular velocity of muscle.

Mean EMG

The ANOVA identified no significant main effect for contraction mode ($F = 1.03$, $p = 0.37$) or angular velocity ($F = 0.87$, $p = 0.47$), and the contraction mode *x* angular velocity interaction ($F = 1.70$, $p = 0.13$) was also not significant. However, a significant main effect for muscle ($F = 63.71$, $p < 0.01$), and, a significant angular velocity *x* muscle ($F = 3.59$, $p < 0.01$) interaction was revealed. Post-hoc analysis highlighted that LG mean EMG was significantly lower ($p \leq 0.01$) than PL and TA at all testing velocities, irrespective of contraction mode, but PL and TA responses were not significantly different between isokinetic speeds ($p \geq 0.10$). There was no contraction mode *x* muscle ($F = 0.62$, $p = 0.65$), or contraction mode *x* angular velocity *x* muscle ($F = 1.03$, $p = 0.42$) interaction. The ANOVA further revealed that there was no significant main effect for limb ($F = 3.90$, $p = 0.07$), nor was there a significant limb *x* contraction mode ($F = 1.21$, $p = 0.32$), limb *x* angular velocity ($F = 0.48$, $p = 0.70$), limb *x* muscle ($F = 1.37$, $p = 0.27$), limb *x* contraction mode *x* angular velocity ($F = 0.86$, $p = 0.53$), limb *x* contraction mode *x* muscle ($F = 1.59$, $p = 0.19$), limb *x* angular velocity *x* muscle ($F = 0.56$, $p = 0.76$), or limb *x* contraction mode *x* angular velocity *x* muscle ($F = 0.61$, $p = 0.83$) interaction.

Peak EMG

There was no significant main effect for contraction mode ($F = 0.26$, $p = 0.77$) or angular velocity ($F = 1.86$, $p = 0.15$), and no significant contraction mode *x* angular velocity ($F = 1.25$, $p = 0.29$) interaction. However, the ANOVA identified a significant main effect for muscle ($F = 57.76$, $p < 0.01$), and a significant angular velocity *x* muscle ($F = 2.56$, $p = 0.03$) interaction. Post-hoc pairwise comparisons highlighted that LG peak EMG responses were significantly lower ($p < 0.01$) than PL and TA at all testing velocities, however there was no significant difference between PL and TA for any velocity ($p \geq 0.12$). There was no significant contraction mode *x* muscle ($F = 1.61$, $p = 0.19$), or contraction mode *x* angular velocity *x* muscle ($F = 0.71$, $p = 0.74$) interaction.

Further analyses revealed no significant main effect for limb ($F = 2.23$, $p = 0.16$), but the limb *x* contraction mode ($F = 4.05$, $p = 0.03$) was statistically significant. Subsequent post-hoc tests demonstrated that peak EMG responses during the ECC_{INV} trials were significantly higher in the non-dominant (110 ± 56 μV; CI; 87-132 μV) limb compared with the dominant (84 ± 34 μV; CI: 73-95 μV, $p = 0.28$, $d = 0.27$) limb, irrespective of angular velocity or muscle. However, there was no significant limb x angular velocity ($F = 0.39$, $p = 0.76$), limb x muscle ($F = 1.75$, $p = 0.19$), limb *x* contraction mode *x* angular velocity ($F = 0.97$, $p =$

0.45), limb *x* contraction mode *x* muscle ($F = 1.77$, $p = 0.15$), limb *x* angular velocity *x* muscle ($F = 0.55$, $p = 0.77$), or limb *x* contraction mode *x* angular velocity *x* muscle ($F = 1.19$, $p = 0.29$) interaction.

iEMG

The ANOVA identified significant main effects for contraction mode ($F = 33.20$, $p < 0.01$), angular velocity ($F = 38.93$, $p < 0.01$) and muscle ($F = 52.90$, $p < 0.01$), along with a significant contraction mode *x* muscle ($F = 12.78$, $p < 0.01$), and angular velocity *x* muscle ($F = 14.34$, $p < 0.01$) interaction. Post-hoc analyses following the significant contraction mode *x* muscle interaction showed that LG activation did not significantly differ between contraction types ($p \geq 0.67$). However, PL response was significantly higher during the ECC_{INV} trials (144 ± 95 µV; CI: 104-185 µV) compared with CON_{EV} (56 ± 44 µV; CI: 34-78 µV, $p < 0.01$, $d = 0.51$) and CON_{INV} (56 ± 31 µV; CI: 42-70 µV, $p < 0.01$, $d = 0.53$). Further, TA response was significantly higher in ECC_{INV} (142 ± 78; CI: 115-168 µV) than in CON_{EV} (67 ± 41 µV; CI: 48-86 µV, $p < 0.01$, $d = 0.45$) and CON_{INV} (72 ± 49 µV; CI: 47-96 µV, $p < 0.01$, $d = 0.42$). Additional findings demonstrated that iEMG was not significantly different ($p \geq 0.82$) between the TA and PL during all contraction modes, but both were significantly greater ($p \leq 0.01$) than LG for all contraction types. There was no significant contraction mode *x* angular velocity *x* muscle ($F = 0.44$, $p = 0.77$) interaction.

Pairwise comparisons subsequent to the ANOVA revealing a significant angular velocity *x* muscle interaction ($F = 14.34$, $p < 0.01$), showed that iEMG was significantly lower ($p < 0.01$) in the LG compared with the PL and TA for all isokinetic testing velocities, irrespective of contraction mode. However, there was no significant difference ($p \geq 0.33$) between PL and TA at all speeds. The main effect for limb was not significant ($F = 2.84$, $p = 0.12$), nor was the limb *x* contraction mode ($F = 2.82$, $p = 0.08$), limb *x* angular velocity ($F = 0.68$, $p = 0.57$), limb *x* muscle ($F = 1.86$, $p = 0.18$), limb *x* contraction mode *x* angular velocity ($F = 0.43$, $p = 0.86$), limb *x* contraction mode *x* muscle ($F = 2.20$, $p = 0.08$), limb *x* angular velocity *x* muscle ($F = 0.49$, $p = 0.81$) or limb *x* contraction mode *x* angular velocity *x* muscle ($F = 0.56$, $p = 0.87$) interaction.

6.4 Discussion

Strength deficits have been implicated as an intrinsic, modifiable risk factor for ankle injury, a prevalent injury in ballet (Willems et al., 2005; Smith et al., 2015). Contrary to previous research in ballet (Thomas and Parcell, 2004; Kenne and Unnithan, 2008), the aim of the current investigation was to quantify ankle eversion and inversion strength, which is fundamental to the common mechanism of injury (Fox et al., 2008), and also, the kinematic demands of ballet-specific jump landing tasks as demonstrated in study 1. The isokinetic strength testing protocol was supported by the sagittal and coronal plane angular displacement and velocity data obtained during study 1, thereby demonstrating greater specificity in research design, and synergy between analysis tools. A more comprehensive profiling of ankle strength in female ballet dancers was evident via inclusion of additional metrics beyond peak torque, that considered strength capacity across the entire torque-velocity curve.

Potential strength imbalances were quantified in respect to contraction mode and movement speed. The significant contraction mode *x* angular velocity interaction demonstrated that ECC_{INV} strength was significantly greater than CON_{EV} and CON_{INV} at all but the slowest angular velocity, with implications for DCRs. This observation is consistent with the classic force-velocity profiles comprising each contraction type, in that concentric strength typically reduces as a product of increasing angular velocity, whereas eccentric strength remains relatively consistent throughout (Cress, Peters and Chandler, 1992). The higher values observed for ECC_{INV} at the greatest angular velocities – which arguably have better functional relevance based on the kinematic findings obtained in study 1 – may be crucial in preventing the inverted foot alignment mechanism common ankle injury incidence (Kaminski et al., 2003). This is particularly important for ballet dancers given that a choreographed routine requires completion of a succession of techniques with a rapid transition up on landing. For example, kinematic analysis of jeté step and sissonne PDB manoeuvres of study 1 demonstrated plantarflexion and a shift towards inversion at the latter stages of the stance phase. Hence, the greater ECC_{INV} at higher speeds observed in this female dancer cohort may help to reduce injury risk when performing these tasks, and indeed, other transitional movements within a routine, and may reflect a chronic training adaptation to the demands of ballet.

The FR metric provides insight on the profile of the strength curve, with higher values indicative of the ability to maintain > 85% of PT for a greater range of motion. In this study, FR was defined at 85% of PT based on observations that a 15% reduction in PT increases injury risk (Croisier et al., 2008). Although this data is not available in previous ballet research, consideration of the FR metric in isokinetic strength analyses may prove vital in identifying markers for injury. At the slower velocities ($30°·s^{-1}$ and $60°·s^{-1}$), ECC_{INV} FR was higher than both concentric modes. However, an inverse trend was established in the contraction mode *x* angular velocity interaction at faster velocities ($90°·s^{-1}$ and $120°·s^{-1}$), whereby CON_{EV} and CON_{INV} FR decreased marginally relative to the significant reductions demonstrated for ECC_{INV}. In accordance with ankle injury aetiology (Skazalski et al., 2017) and the ankle kinematic responses to ballet movement highlighted in study 1, the notable decline in ECC_{INV} FR at increased velocities may have implications on injury susceptibility when executing ballet-specific locomotion. Lower ECC_{INV} may compromise the ability to resist inversion forces thereby proliferating injury risk (Fox et al., 2008), especially during the transition between connecting elements of a routine. Rather than using PT – which provides a single strength value for a pre-determined range of motion – in the pursuit of identifying markers for injury, professionals with an injury reduction focus may benefit from the FR metric during isokinetic strength testing (Croisier et al., 2008). The significant reductions in ECC_{INV} FR at angular velocities exceeding $60°·s^{-1}$ provides a focus for subsequent strength training interventions. Even if PT and the maxima of strength curve is unchanged, a reduction in FR suggests a decrease in strength away from the single joint angle defined as APT. Practically, a dancer would benefit from high PT and FR since performance demands will move through an angular range at the ankle. However, it should be acknowledged that a decrease in FR across all contraction modes at higher velocities may be indicative of a limited isokinetic phase as the dynamometer crank arm accelerates to higher speeds over a relatively small range of motion. Torque may indeed be maintained at 85% of PT outside the isokinetic range, but the restricted focus on the isokinetic data curtails the FR metric. The range of movement for ankle eversion and inversion is smaller than knee flexion and extension (Eustace, Page and Greig, 2017), and thus, FR values appear to be joint specific and should be interpreted with this in mind. Moreover, direct comparisons of FR between studies may only be achieved when uniform methodological designs are used.

Conventional DCRs are derived from PT values without any consideration of the angle at which PT is achieved, and thus, limit an understanding of how strength changes as a function of angle (El-Ashker et al., 2017). Data from the current investigation demonstrated that peak

CON_{EV} and CON_{INV} torque was achieved at ~18° (2° of inversion) and ~17° (3° of eversion), thereby representing a relatively neutral foot alignment over the 40° range of motion. Using the kinematic data of study 1, peak CON_{EV} and CON_{INV} strength occurs within the first 25% of the stance phase for all movements. The early achievement of peak concentric strength during the landing phase of ballet techniques, coupled with increased FR in these contraction modes, may be crucial in adopting and maintaining a stable joint position in eversion to manage vGRFs (Bonnel et al., 2010). ECC_{INV} PT occurred significantly later at ~27°, representing a 7° position of inversion. Whilst the land and hold manoeuvres of study 1 presented an everted foot position, the 7° displacement in inversion for peak eccentric strength has important clinical implications for the land and transition tasks. With a progression towards inversion observed during the jeté and sissonne PDB trials, peak eccentric strength in an inverted foot configuration may help to reduce the risk of injury during ballet performance.

Previous literature has failed to quantify the angle of peak torque for ankle eversion/inversion isokinetic strength ratios, preventing direct comparison. However, studies examining strength parameters at the knee joint have demonstrated that APT varies between concentric and eccentric modes of contraction and across a range of angular velocities (Cohen et al., 2015; Eustace, Page and Greig, 2017). The evident inconsistencies for APT from the current study and indeed other investigations, raises questions over the value of traditional PT-derived strength ratios and supports the inclusion of AST assessed across a number of angular velocities. The greater PT but smaller FR in ECC_{INV} at higher velocities has implications for functional performance and the strategies deployed in training and/or rehabilitation. The inverted configuration in ECC_{INV} – approximately 7° – compared with concentric modes may serve as a protective mechanism against ankle injury during ballet. These inferences remain unsubstantiated without other research on which to compare with, and further investigation into the importance of peak ECC_{INV} angular displacement and injury susceptibility is required.

Findings from the study revealed significant main effects for contraction mode, angle, and angular velocity on AST, whilst significant contraction mode *x* angle and contraction mode *x* angular velocity interactions were also revealed. Data for AST and corresponding control ratios were significantly higher for ECC_{INV} than CON_{EV} and $CON_{INV,}$ with more profound differences observed at the latter ranges of the movement, and with increasing angular velocity. This finding may be attributed to both the force-velocity and force-angle relationships between contraction modes. CON_{EV} and CON_{INV} strength portray a quadratic trend, in which PT is achieved at approximately midpoint of the movement, whereas ECC_{INV} PT is achieved towards end range. The use of angle-specific torque is sensitive to the changes in strength at various positions within a movement. Resultant DCRs may be used in the screening of performers towards injury reduction, and in the management of injury during rehabilitation. For example, in the current study, whilst decreases in CON_{INV} strength near to full ankle inversion were exacerbated as angular velocity increased, ECC_{INV} was relatively consistent. Eccentric inversion dominance at the end ranges of movement and at higher velocities may indeed reduce the likelihood of ankle injury when executing the jump landing, cutting manoeuvres of a ballet routine.

Main effects for limb dominance were investigated for all isokinetic outcome measures in consideration of the aesthetic demand for movement symmetry within choreographed routines. There was no evidence of bilateral asymmetry in any of the isokinetic ankle strength measures, reflecting the kinematic and kinetic findings from study one, and the EMG responses from study 2. The primary responsibility of the ankle joint is to attenuate the resultant mechanical loads imposed by movement, with strength playing a significant role (Michael et, 2008). With over half of jumps during a typical routine comprising a unilateral landing (Mertz and Docherty, 2012), strength symmetry between limbs is desirable to minimise a greater loading and ensuing tendency to sustain injury in a particular side. Emphasising symmetry in limb control within dance training may be crucial in developing equal bilateral strength and reducing injury risk, and, may partially explain the non-significant differences in the cohort of female dancers used in this investigation. Ballet epidemiology literature is not available to critically discuss this findings in relation to injury risk, prompting a need for further research in this field.

EMG Responses

Neuromuscular activity during the isokinetic strength protocol was evaluated given the inter-relationship with force production (Roberts and Gabaldon, 2008). The first key observation from EMG analysis showed that contraction mode had no significant influence on mean or peak EMG response. Whether driving movement or arresting momentum during a deceleration/landing task, coordinated co-activation of all musculature surrounding a joint in all modes of contraction is required to provide dynamic joint stabilisation (Kaminski and Hartsell, 2002). Although joint movement plane is restricted to a single vector during isokinetic strength testing, the evident balance in EMG response demonstrated in this population has important kinematic implications when performing ballet-specific movements. The eversion and inversion mechanism common to the jump landing tasks of study 1 emphasises the importance of concerted neuromuscular control to attenuate GRFs, and, to facilitate a smooth transition into a proceeding technique (Withrow et al., 2008). A second key finding concerned the significant angular velocity *x* muscle interaction. Irrespective of contraction mode, muscular activity in LG was significantly lower than PL and TA. As expected, the lower activation in LG is symptomatic of functional kinesiology. Although the foot attachment was set in partial plantarflexion, the isokinetic strength protocol moved through eversion and inversion reflecting in greater PL and TA response. The significant limb *x* contraction mode interaction highlighted that peak EMG responses were significantly greater in the non-dominant limb compared with the dominant limb during the ECC_{INV} trials. Though difficult to qualify without a significant interaction involving muscle, the significant differences between limbs may be due to the greater level of PL activity observed in the non-dominant limb during the ECC_{INV} trials. Bilateral asymmetry in EMG response was not observed in measures of mean or iEMG, and therefore, inferences regarding a potential asymmetry in neuromuscular control between limbs should be made with caution.

EMG analysis was conducted to strengthen the understanding on the muscular strategy used during isokinetic strength testing, but also, to enable a comparison between the EMG responses obtained during study 2. During the ballet-specific movement battery of study 2, gross mean and peak EMG responses approximated 140 µV and 336 µV respectively. Mean and peak EMG responses during the isokinetic strength testing protocol were ~53 µV and ~97 µV, and thus, were 3-4 times lower than the movement battery. Various factors may qualify this observation. Firstly, performing ballet-specific manoeuvres requires a coordinated and simultaneous input from various joints to maintain correct ballet posture and to provide stability during landing. The dynamic profile of ballet comprises elevated GRFs, which in turn, places a greater demand on the neuromuscular system in managing the kinetic and kinematic response. Secondly, there was greater LG and PL activity in the movement battery compared with the isokinetic strength protocol. Although LG responses were lowest for both assessments, the ballet-specific tasks demanded more plantarflexion during initial contact, and also during the transitional component of the jeté and sissonne PDB. Regarding PL activation, the higher levels observed in the ballet-specific movement battery of study 2 may be explained by the magnitude of joint displacement in the coronal plane. A position of maximal eversion (20°) during the isokinetic strength testing protocol was equalled or exceeded by all ballet-specific tasks except for the échappé, which likely requires a corresponding increase in PL response. Further, the everted joint position during the strength evaluation was only held for a very limited period, whereas eversion was maintained for around 75% of the stance phase during the land and hold techniques during study. This also supports the exclusion of a general EMG normalisation method from the current thesis, as normalising response may indeed need to be task-specific.

Given the lack of existing research on which to draw comparisons, caution is advised when generalising the findings of the current study beyond the specific experimental design employed. No other study has investigated the ankle eversion and inversion strength in female ballet dancers, nor is there any other research quantifying neuromuscular responses to isokinetic movement. Isokinetic testing protocols have some inherent methodological constraints, which should be considered when developing the data collection paradigm. The joint range of motion and joint angular velocities used in the current study are close to the physiological capabilities of the ankle when tested in this restricted state. Pilot testing highlighted that no isokinetic phase was determined at angular velocities of $\geq 150°\cdot s^{-1}$, and the range of motion is prescribed using passive movement of the joint. Consequently, this passive manipulation of the joint within an isokinetic testing paradigm may not reflect the physical capacity of the joint during active movement. The present study focused on ankle eversion/inversion strength given its mechanical associations with injury. However, a strength profile of dancers may include plantar/dorsiflexion strength in light of kinematic analyses highlighting ankle injuries to occur in neutral (Fong et al., 2009) and dorsiflexed positions (Kristianslund, Bahr, and Krosshaug, 2011). Kinematic analyses of injury incidence or ballet-specific movements may inform bespoke isokinetic testing protocols but must also account for physical the limitations of the ankle during such assessments. Further research is required in the associations between isokinetic metrics and injury incidence, which may inform a threshold for the calculation of FR. Contemporary analysis metrics that delve beyond the highest value of a strength curve are advocated. Data collection ought to utilise the capacity of the isokinetic dynamometer to measure net joint torque at predetermined angles and angular velocities to provide a screening battery of greatest functional relevance to the sport.

6.5 Conclusion

This is the first study to consider the eversion/inversion strength of female ballet dancers, with the protocol supported by the ankle kinematic data obtained in study 1. The current investigation also represents the first to evaluate EMG responses to isokinetic strength testing, and to compare functional EMG responses with EMG capacity during maximal testing. The isokinetic strength profile of ballet dancers in this study illustrated that ECC_{INV} strength is maintained over a range of angular velocities, compared with reductions in CON_{EV} and CON_{INV} strength as movement speed increases. Specifically, ballet dancers appear to be ECC_{INV} at angular velocities of $60°\cdot s^{-1}$ and beyond, and for all angular displacements, which has important implications with regards to the ankle joint angular velocities observed in study 1. Beyond the singular peak of the torque-angle curve, ECC_{INV} had greater FR at velocities $< 90°\cdot s^{-1}$ compared with concentric modes, which may indicate a protective mechanism for injury. Female ballet dancers demonstrate bilateral symmetry in strength, reflecting the kinetic and kinematic observations of study 1, and the EMG responses of study 2. This observation may be attributed to both appropriate training interventions from an early age , and, regular exposure to the asymmetric movement patterns of ballet that facilitates bilateral strength development. The interactions between joint angle, angular velocity and contraction mode during isokinetic testing offer an attractive method in screening of dancers towards understanding the occurrence of injury, and, in the pursuit of performance excellence. Despite the demonstrable novelty in the development of a functionally relevant isokinetic testing protocol based on the kinematic data obtained in study 1, the use of isokinetic dynamometers restrict biomechanical analyses to laboratory conditions. In addition, dynamometer constraints and ethical boundaries in testing prohibit an evaluation of true ballet movement. Alternative methods for investigating the biomechanics of ballet in field-based settings offer greater ecological validity. Neuromuscular activity informs strength, and therefore EMG may be a preferable tool for in-vivo measurement during performance monitoring. Portability in analysis methods provides a means of conducting biomechanical investigations of ballet, rather than of ballet dancers.

Chapter 7. Study four: Quantifying Mechanical Load during Ballet-specific Movements using Tri-axial Accelerometry

7.1 Introduction

In studies 1 and 2, a battery of ballet-specific jump landing tasks was used to investigate kinetic, kinematic and neuromuscular responses, with resultant data sensitive to task characteristics regarding the varying planar demands comprising each movement. Study 3 evaluated isokinetic strength capacity during a testing protocol supported by the joint kinematic data obtained in study 1, and compared muscular responses with the movement battery of study 2. Although greater specificity towards ballet performance was evident in methodological design, these analyses were conducted in laboratory settings, with 3D motion capture systems, force plates, and isokinetic dynamometers lacking portability beyond this environment. Hence, alternative biomechanical analysis methods that are capable of sophisticated in-vivo assessments of the same ballet movements, but in authentic dance settings, will help to enhance ecological validity in biomechanics research and advance current practice. Contemporary developments in motion tracking technologies such as Global Positioning Systems (GPS) may offer a pathway to bridge laboratory- and field-based investigations in biomechanics. in

GPS-based wearable tracking devices enable comprehensive assessments of human movement during sport via recording of numerous distance- and velocity-focused metrics (Jennings et al., 2010; Boyd, Ball and Aughey, 2011; Dunbar et al., 2014). To that end, GPS technology has contributed to a considerable understanding of the physical characteristics pertaining to sport, particularly in team invasion modes (Gabbett, Jenkins and Abernethy, 2010; Cormack et al., 2013; Scott et al., 2013; White and McFarlane, 2013). Whilst the real-time data gathered assists in the physical preparation of athletes toward the rigours of performance, the low sampling frequency (1-10 Hz) of GPS-based analyses returns questionable validity and reliability to accurately assess short, high-intensity movements (Nicolella et al., 2018). Comparatively less attention has been directed to the implications of sport imposing loads on the musculoskeletal system from a biomechanical perspective. Most commercially available GPS units house inertial measurement units such as tri-axial accelerometers with a capacity to record data at 100 Hz (Chambers et al., 2015). Quantifying three-dimensional accelerations at this increased sampling frequency is more representative of the 3D motion capture system used in study 1. Moreover, the tri-planar evaluation of movement using accelerometry mirrors traditional force plate analysis, but its utility suits the explosive, multi-directional movement profile inherent to ballet (Twitchett, Koutedakis and Wyon, 2009). The calculation of acceleration in the three movement planes is subsequently used to quantify mechanical load (Waldron et al., 2011). The most frequently reported accelerometer metric is total PlayerLoad™ (Figure 7.1) which is defined as a modified vector magnitude, expressed as the square root of the sum of the squared instantaneous rate of change in acceleration in each of the three vectors (Boyd, Ball and Aughey, 2011; Cormack et al., 2013; Barron et al., 2014).

$$\text{Player load} = \sqrt{\frac{\left(a_{y1} - a_{y-1}\right)^2 + \left(a_{x1} - a_{x-1}\right)^2 + \left(a_{z1} - a_{z-1}\right)^2}{100}}$$

Figure 7.1. The PlayerLoad™ equation.

Where a_y is forward (anterior-posterior acceleration), a_x is sideways (medial-lateral acceleration, and a_z is vertical acceleration.

The PlayerLoad™ vector representation is comparable to the product of adding each vector from a GRF signal to represent accumulated ground reaction force, a method which is flawed and potentially hinders the interpretation of how total body load is achieved. For example, a battery of tasks may elicit similar total PlayerLoad™ responses, irrespective of the disparity in multi-planar characteristics specific to each movement. The rationale and value of uni planar consideration in load assessment has been emphasised in previous research (Greig and Nagy, 2017). The findings of this study highlighted increased contributions of mediolateral accelerations to total PlayerLoad™ in the lumbar spine of cricket fast bowlers, reflecting both the fast bowling action and injury epidemiology observations. To that end, quantifying the relative planar contributions in accelerometer-based investigations is advocated. The sign convention issue in squaring each uni-axial vector prohibits an understanding regarding the direction of movements completed in each plane (i.e. forwards/backwards in the *ay* plane). Refuting the differences in anterior and posterior motion, and the relative contributions to total PlayerLoad™ considerably restricts the level of analysis towards examining movement quality. Moreover, it may potentially mask issues in uni-planar kinematics such as greater lateral deviations of the body, with implications for injury risk.

Despite its use in other sporting modalities (Cummins et al,. 2013), accelerometer-derived measures of mechanical load during ballet movement are scarce (Brogden et al., 2018). The aim of the current investigation therefore, was to quantify the mechanical responses to the same ballet-specific jump landing tasks of studies 1 and 2 using tri-axial accelerometry. Based on the kinetic and kinematic findings of the previous experimental chapters, the principles of accelerometry should be sensitive to the planar demands comprising each task, with resultant accelerometer-derived load profiles differentiating the movements. The study further aimed to quantify the uni-axial contributors to overall movement for each task, and highlight within-planar disparities in movement direction. Potential findings may highlight the sensitivity of accelerometry in detecting multi-planar differences in ballet motion, which may support the use of accelerometry in contemporary ballet load monitoring strategies.

Experimental research question:

1) Is accelerometry sensitive to the varying planar characteristics comprising ballet-specific movement?

2) Does accelerometry offer an efficacious means of quantifying mechanical load responses to ballet specific movements?

7.2 Methodology

Participants

The participants for this experimental study involved the same cohort of female dancers used in studies 1, 2 and 3. However, a retrospective analysis of the data revealed that accelerometer data did not record for 2 participants leaving a sample of 12 (Age, 19.42 ± 1.68 yrs; Height, 1.64 ± 0.06 m; Body mass, 62.13 ± 8.62 kg). All participants provided written informed consent, conformed with the inclusion criteria outlined in the General Methodology (Chapter 3), and completed the necessary pre-screening protocols also contained in Chapter 3.

Procedures

Each participant was required to visit the Biomechanics Laboratory on one occasion to complete the experimental trial. The warm-up procedures and the experimental trials were identical to those completed in study 1.

Tri-axial accelerometry

Each participant was fitted with a GPS device (MinimaxX S4, Catapult Innovations, Melbourne, Australia) which located at the mid-point between the scapula (approximating the 7th cervical vertebrae). To reduce extraneous device movement, the unit was placed within the neoprene pouch of the manufacturer provided vest. The integrated accelerometer (Kionix KX94, Kionix, Ithaca, New York, USA) was used to quantify individual planar accelerations, with data sampling at 100 Hz. Each participant was afforded time to familiarise with the feel of the GPS unit to prevent any compromised or unnatural movement during the experimental trials.

Data Processing

Accelerometer data was downloaded in the corresponding GPS software (Catapult Sprint V5.1.4) for processing. A period for each movement block, and subsequently each jump landing trial, was delineated to allow relevant accelerometer metrics to be extracted. Accumulated total PlayerLoad^{TM} (PL_{TOTAL}) was calculated for each ballet-specific jump landing task. The relative uni-axial planar contributions to total load for each vector (mediolateral - $PL_{ML\%}$; anteroposterior – $PL_{AP\%}$; vertical – $PL_{V\%}$) were subsequently calculated. Reliability analysis was conducted for PL_{TOTAL} and the individual contributing planar indices, with corresponding values contained in Table 7.1 and interpreted according to the thresholds defined in Chapter 4. Raw acceleration traces were exported to Excel (Microsoft Corporation, Washington, USA) for further analysis. For each jump landing task, peak acceleration values in each planar direction; mediolateral (ACC_{ML}), anteroposterior (ACC_{AP}), vertical (ACC_{V}) were determined by using the max and min values. Max values refer to forwards, right and upwards movement for the medial-lateral, anterior-posterior and vertical planes respectively; whereas min values refer to backwards, left and downwards motion. Resultant values were used to determine an acceleration symmetry index (SI) in each direction. An index value of less than 1 indicates greater accelerations in backwards, left and downwards motion, whereas values of greater than 1 represent greater forwards, right and upwards accelerations.

Table 7.1. Measures of reliability (ICC – (95% CI)) for PlayerLoad^{TM} data across the movement battery.

		Jump-Landing Manoeuvre						
Variable	**Limb**	**Jeté**	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
PL_{AP}	Dominant	0.98 (0.94-0.99)	0.98 (0.94-0.99)	0.95 (0.86-0.98)	0.91 (0.76-0.98)	0.94 (0.83-0.98)	0.93 (0.80-0.98)	0.94 (0.83-0.98)
	Non-Dominant	0.98 (0.93-0.99)	0.99 (0.97-1.00)	0.91 (0.75-0.97)	0.95 (0.86-0.99)	0.95 (0.86-0.99)	0.83 (0.55-0.95)	0.97 (0.91-0.99)
PL_{ML}	Dominant	0.97 (0.92-0.99)	0.99 (0.97-1.00)	0.88 (0.68-0.97)	0.72 (0.25-0.92)	0.94 (0.94-0.98)	0.96 (0.89-0.99)	0.91 (0.75-0.97)
	Non-Dominant	0.97 (0.93-0.99)	0.98 (0.94-0.99)	0.88 (0.70-0.97)	0.91 (0.75-0.97)	0.97 (0.92-0.99)	0.96 (0.88-0.99)	0.93 (0.81-0.98)
PL_V	Dominant	0.97 (0.93-0.99)	0.98 (0.95-1.00)	0.96 (0.88-0.99)	0.91 (0.74-0.97)	0.97 (0.90-0.99)	0.99 (0.97-1.00)	0.93 (0.80-0.98)
	Non-Dominant	0.99 (0.97-1.00)	0.99 (0.97-1.00)	0.96 (0.88-0.99)	0.97 (0.91-0.99)	0.98 (0.95-0.99)	0.98 (0.96-1.00)	0.91 (0.76-0.97)
PL_{TOTAL}	Dominant	0.98 (0.95-1.00)	0.99 (0.97 -1.00)	0.98 (0.94-0.99)	0.89 (0.70-0.97)	0.97 (0.92-0.99)	0.95 (0.86-0.98)	0.94 (0.83-0.98)
	Non-Dominant	0.99 (0.97-1.00)	0.99 (0.98-1.00)	0.96 (0.88-0.99)	0.98 (0.93-0.99)	0.98 (0.94-0.99)	0.91 (0.76-0.97)	0.95 (0.86-0.99)

PL_{AP}, anteroposterior PlayerLoad^{TM}; PL_{ML}, mediolateral PlayerLoad^{TM}; PL_V, vertical PlayerLoad^{TM}.

Statistical Analysis

Statistical tests on all data parameters were conducted using a statistics software package (IBM SPSS Statistics V25.0, IBM, Armonk, New York, USA). Descriptive data (mean ± σ) are presented for each PlayerLoad^{TM} and acceleration-derived metric across the battery of movement tasks. The distribution of data around the mean was quantified using Q-Q plots, histograms and a Shapiro-Wilk test. With the data normality assumptions verified, separate repeated measures ANOVAs were conducted for each accelerometer metric. A 7 x 2 (movement x limb) model was used to investigate main effects and interactions for PL_{TOTAL}, a 7 x 3 x 2 (movement x plane x limb) for uni-planar contributions, and a 7 x 3 x 2 x 2 (movement x plane x direction x limb) for within-planar accelerations. Wherever a statistically significant difference was identified, post-hoc pairwise comparisons with a Bonferroni correction factor were employed to highlight which variables were significantly different from one another. Confidence Intervals (CI) with a 95% threshold and Cohen's *d* effect sizes (small; 0.20-0.49; moderate, 0.50-0.79; large, ≥ 0.80) were also presented. A 0.05 alpha level was used to determine statistically significant differences.

7.3 Results

PlayerLoad^{TM} Responses

Figure 7.2 displays the influence of movement task and laterality on resultant accumulated load and the corresponding uni-axial indices of load. The ANOVA revealed a significant main effect for movement on PL_{TOTAL} ($F = 6.09$, $p = 0.02$), PL_{ML} ($F = 9.48$, $p < 0.01$) and PL_{V} ($F = 6.39$, $p = 0.02$), but not for PL_{AP} ($F = 3.47$, $p = 0.06$). However, there was no significant main effect for limb on PL_{TOTAL} ($F = 0.02$, $p = 0.90$), PL_{AP} ($F = 0.01$, $p = 0.94$), PL_{ML} ($F = 0.20$, $p = 0.67$) or PL_{V} ($F = 0.02$, $p = 0.89$), nor a significant movement *x* limb interaction for any load metric ($p \geq 0.54$).

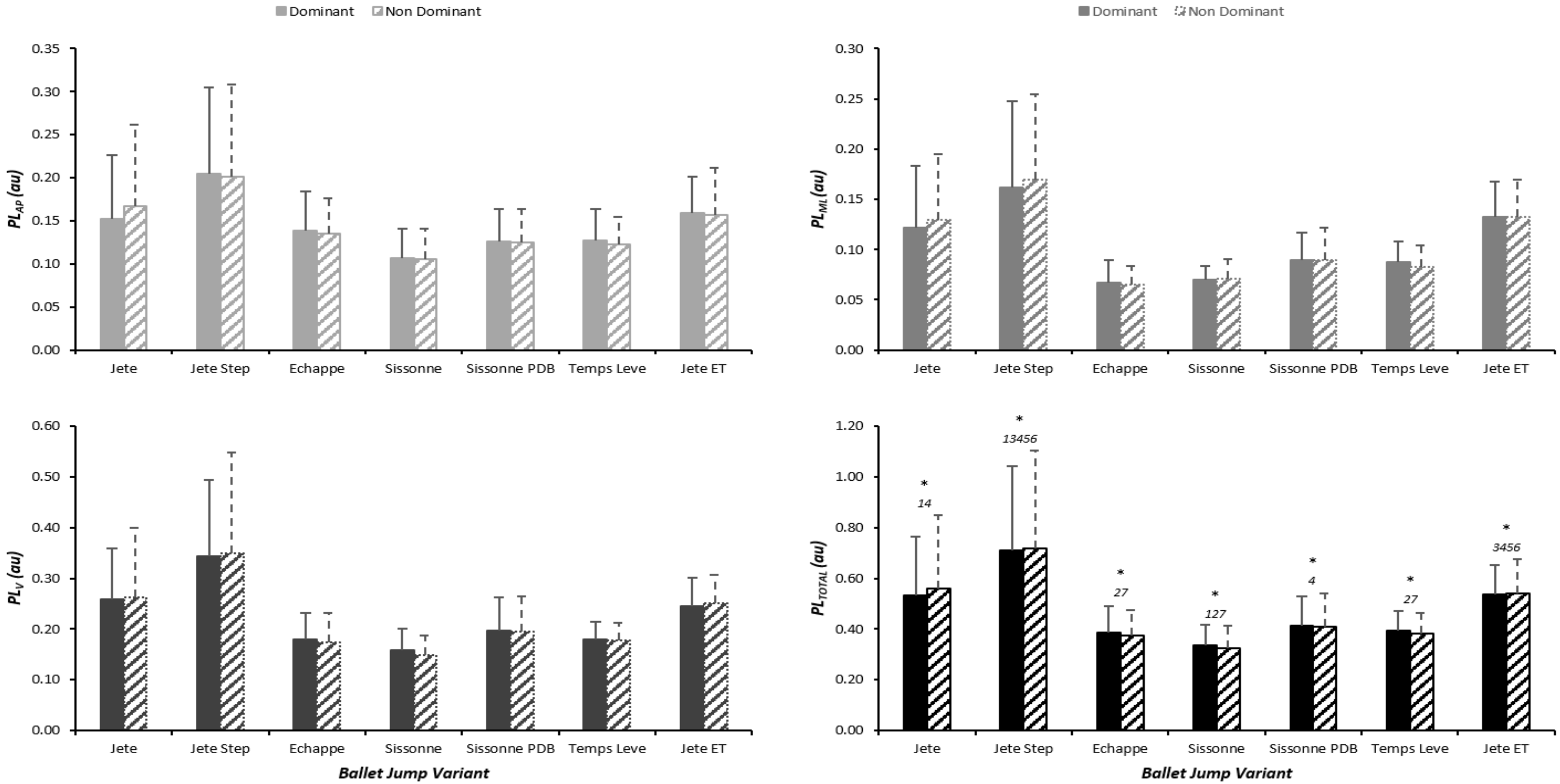

Figure 7.2. Uni-axial and total PlayerLoad^{TM} responses to each ballet-specific movement for the dominant and non-dominant limbs. * Denotes a significant main effect for movement. ^{1234567} signify the pairwise comparisons for Jeté (^{1}) to Jeté ET (^{7}). Corresponding values are mean ± σ.

Post-hoc analyses highlighted that PL_{TOTAL} was significantly higher during the jeté step (0.72 ± 0.36 au; CI: 0.47-0.95) compared with all other movements ($p \leq 0.04$) except for the jeté ET (0.54 ± 0.13 au; CI: 0.46-0.62, p = 0.16). Additional between-movement comparisons revealed that PL_{TOTAL} during the jeté ET was significantly greater than the échappé (0.38 ± 0.10 au; CI: 0.31-0.45, p = 0.01, *d* = 0.57), sissonne (0.33 ± 0.09 au; CI: 0.28-0.39, p < 0.01, *d* = 0.68), sissonne PDB (0.41 ± 0.13 au; CI: 0.33-0.49, p = 0.02, *d* = .045), and temps levé (0.39 ± 0.08, p = 0.01, *d* = 0.57). Finally, PL_{TOTAL} was significantly greater during the jeté (0.55 ± 0.26 au; CI: 0.37-0.72) and sissonne PDB compared with the sissonne ($p \leq 0.04$, *d* = 0.37-0.49).

Pairwise comparisons showed that PL_{ML} during the jeté ET (0.13 ± 0.04 au; CI: 0.11-0.16) was significantly greater than the échappé (0.07 ± 0.02 au; CI: 0.05-0.08, p < 0.01, *d* = 0.69), sissonne (0.07 ± 0.02 au; CI: 0.06-0.08, p < 0.01, *d* = 0.69), sissonne PDB (0.09 ± 0.03 au; CI: 0.07-0.11, p = 0.03, *d* = 0.49), and temps levé (0.09 ± 0.02 au; CI: 0.07-0.10, p = 0.02, *d* = 0.53). There were no further significant between-movement differences for PL_{ML} ($p \geq 0.05$). Additional post-hoc analyses for the PL_V metric highlighted that PL_V was significantly greater during the jeté ET (0.25 ± 0.06 au; CI: 0.21-0.28) compared with the sissonne (0.16 ± 0.04 au; CI: 0.13-0.18 , p < 0.01, *d* = 0.66). No other significant between-movement differences for PL_V were identified ($p \geq 0.13$).

Relative Uni-axial Contributions

The influence of movement task and laterality on the relative planar contributions to PL_{TOTAL} is summarised in Table 7.2. The ANOVA revealed a significant main effect for uni-axial contribution (F = 168.85, p < 0.01), with post-hoc analyses revealing that PL_V (47.21 ± 4.05 %; CI: 45.14-49.28) represented a significantly greater proportion of PL_{TOTAL} compared with PL_{AP} (30.80 ± 3.95 %; CI: 29.12-32.80, p < 0.01, *d* = 0.90) and PL_{ML} (21.99 ± 3.61 %; CI: 21.00-23.52, p < 0.01, *d* = 0.96) respectively. Contributions in PL_{AP} were significantly greater than PL_{ML} (p < 0.01, *d* = 0.76). The movement *x* uni-axial contribution interaction was also significant (F = 6.20, p < 0.01), and Table 7.1 highlights where the uni-axial contributions differ between movements. There was no significant main effect identified for limb on $PL_{AP\%}$ (F = 7.72, p < 0.01), $PL_{ML\%}$ (F = 1.06, p = 0.33), or $PL_{V\%}$ (F = 3.23, p = 0.10), nor was there any significant limb *x* movement (F = 2.27, p = 0.06), limb *x* uni-axial contribution (F = 1.11, p = 0.35), or limb *x* movement *x* uni-axial contribution (F = 0.89, p = 0.56) interactions.

Table 7.2. The effects of ballet task and limb on uni planar contributions to PL_{TOTAL}. Corresponding values are mean ± σ.

Variable	Limb	Ballet Jump Variant						
		Jeté	Jeté step	Échappé	Sissonne	Sissonne PDB	Temps levé	Jeté ET
$PL_{AP\%}$ (CI)	Dominant	28.10 ± 3.23$^{\delta 3}$	28.51 ± 3.07$^{\delta 35}$	35.61 ± 5.49$^{\delta 127}$	31.31 ± 4.67$^{\delta}$	30.39 ± 3.05$^{\delta}$	31.81 ± 4.16$^{\delta 2}$	29.67 ± 4.31$^{\delta 7}$
		(25.93-20.28)	(26.45-30.57)	(31.93-39.31)	(28.17-34.44)	(28.34-32.43)	(29.02- 34.61)	(26.77-32.57)
	Non-Dominant	29.25 ± 3.71$^{\delta 3}$	27.82 ± 3.20$^{\delta 3}$	35.75 ± 5.52$^{\delta 127}$	31.90 ± 3.25$^{\delta}$	30.62 ± 3.60$^{\delta}$	31.85 ± 3.11$^{\delta}$	28.57 ± 4.90$^{\delta 7}$
		(26.76-31.75)	(25.67-29.97)	(32.05-49.47)	(29.72-34.09)	(28.20-33.04)	(29.76-33.94)	(25.27-31.86)
$PL_{ML\%}$ (CI)	Dominant	22.53 ± 3.14^{3}	22.36 ± 3.53^{3}	17.56 ± 3.74^{127}	21.25 ± 3.52	22.00 ± 3.64	22.32 ± 4.90	24.55 ± 3.00^{3}
		(20.42-24.64)	(19.99-24.73)	(15.04-20.07)	(18.89-23.62)	(19.56-24.45)	(19.03-25.61)	(22.53-26.57)
	Non-Dominant	23.47 ± 4.57^{3}	23.72 ± 4.35^{3}	17.98 ± 4.47^{127}	22.13 ± 2.61	21.78 ± 3.11	21.56 ± 2.23	24.72 ± 3.74^{3}
		(20.40-26.54)	(20.80-26.65)	(14.98-20.99)	(20.37-23.88)	(19.69-23.87)	(20.06- 23.06)	(22.20-27.23)
$PL_{V\%}^{*}$ (CI)	Dominant	49.36 ± 3.10	49.14 ± 3.59	46.82 ± 4.97	47.44 ± 4.17	47.62 ± 4.99	45.86 ± 3.44	45.78 ± 4.53
		(47.28-51.44)	(46.72-51.55)	(43.48-50.16)	(44.64-50.24)	(44.27-50.97)	(43.55-48.17)	(42.73-48.83)
	Non-Dominant	47.28 ± 3.17	48.46 ± 4.80	46.26 ± 5.54	45.97 ± 2.84	47.60 ± 4.32	46.59 ± 4.35	46.72 ± 2.90
		(45.15-49.41)	(45.24-51.69)	(42.54-49.98)	(44.07-47.88)	(44.70-50.50)	(43.67-49.51)	(44.77-48.66)
Total		100	100	100	100	100	100	100

$PL_{AP\%}$, anteroposterior PlayerLoad™ contribution; $PL_{ML\%}$, mediolateral PlayerLoad™ contribution; $PL_{V\%}$, vertical PlayerLoad™ contribution. CI, 95% confidence intervals. * denotes that the contribution in PL_V is significantly greater ($p < 0.05$) than PL_{AP} and PL_{ML} for all movements. $^{\delta}$ denotes a significant difference ($p < 0.05$) between PL_{AP} and PL_{ML}. ^{1234567} signifies which movements are significantly difference from each other, with (^{1}) representing the Jeté and (^{7}) the Jeté ET.

Planar Accelerations

Uni-planar accelerations corresponding to each jump landing tasks and for each limb are presented in Figure 7.3. There was no significant main effect for movement on ACC_{AP} (F = 1.45, p = 0.25), but there was for ACC_{ML} (F = 6.80, p = 0.01) and ACC_{V} (F = 7.00, p = 0.01). Post-hoc analyses revealed that ACC_{ML} was significantly lower in the échappé (0.55 ± 0.34 g; CI: 0.42-0.68) compared with the jeté (0.85 ± 0.38 g; CI: 0.70-1.00, p = 0.22, *d* = 0.38), jeté step (0.87 ± 0.38 g; CI: 0.68-1.06, p = 0.02, *d* = 0.41) and the jeté ET (0.93 ± 0.52 g; CI: 0.74-1.13, p = 0.01, *d* = 0.40), irrespective of direction and limb. There were no further significant between-movement differences in ACC_{ML} ($p \geq 0.13$). For ACC_{V}, pairwise comparisons highlighted that ACC_{V} was significantly greater in the jeté step (2.35 ± 0.74 g; CI: 2.01-2.70) compared with the sissonne (1.61 ± 0.38 g; CI: 1.44-1.78, p = 0.04, *d* = 0.53) and temps levé (1.71 ± 0.39 g; CI: 1.51-1.92, p = 0.02, *d* = 0.48). No other significant differences ($p \geq 0.08$) in ACC_{V} were identified between movements.

A further significant main effect was identified for direction on ACC_{AP} (F = 344.05, p = 0.01) and ACC_{V} (F = 259.61, p = 0.01), but not for ACC_{ML} (F = 0.34, p = 0.57). Pairwise analyses showed that posterior accelerations (1.79 ± 0.45 g; CI: 1.63-1.95) were significantly greater than anterior accelerations (0.52 ± 0.29 g; CI: 0.41-0.64, p = 0.01, *d* = 0.83), irrespective of movement and limb. In addition, upwards accelerations (3.11 ± 0.65 g; CI: 2.85-3.37) were significantly greater than downwards accelerations (0.60 ± 0.37 g; CI: 0.44-0.77, p = 0.01, *d* = 0.92), irrespective of movement and limb.

ANOVA also revealed a significant main effect for limb on ACC_{ML} (F = 10.27, p = 0.01), with greater planar accelerations observed in the non-dominant (0.80 ± 0.40 g; CI: 0.69-0.91) compared with the dominant (0.73 ± 0.33 g; CI: 0.64-0.82, p = 0.01, *d* = 0.10) limb, irrespective of movement and direction. However, this effect was not evident for ACC_{AP} (F = 0.26, p = 0.62) or ACC_{V} (F = 1.34, p = 0.27), and there were no significant limb *x* movement ($p \geq 0.30$), limb *x* direction ($p \geq 0.67$), or limb *x* movement *x* direction ($p \geq 0.15$) interaction for any planar accelerations.

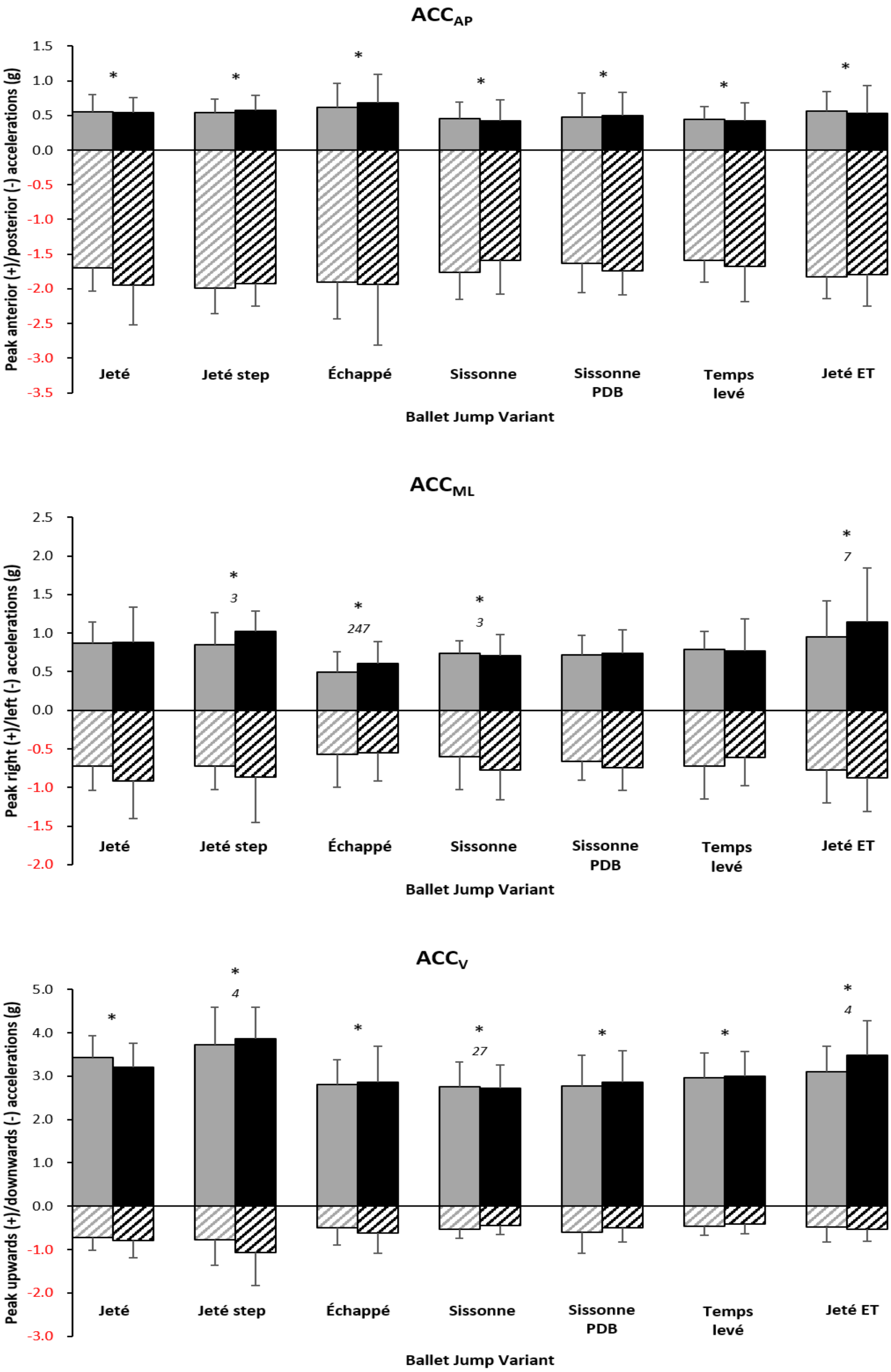

Figure 7.3. Stacked column bar charts representing maximum (solid fill) and minimum (pattern fill) uni-planar accelerations for the dominant (grey) and non-dominant (black) limb for each ballet-specific jump landing task. * indicates a significant main effect for direction. *1234567* signifies which tasks are significantly different from each other, (*1*) is Jeté (*7*) Jeté ET. Corresponding values are mean ± σ.

The movement x plane ($F = 5.05$, $p < 0.01$), movement x direction ($F = 3.13$, $p = 0.01$), plane x direction ($F = 260.59$, $p < 0.01$) and movement x plane x direction ($F = 1.98$, $p = 0.03$) interactions were significant. Post-hoc tests revealed that in the anteroposterior plane, backwards accelerations were significantly greater ($p \leq 0.01$) than forwards accelerations for all movements. Further comparisons found no significant difference ($p \geq 0.30$) between left and right accelerations in the mediolateral plane for all movements, but upwards accelerations were significantly greater ($p \leq 0.01$) than downwards accelerations during all tasks.

In the anteroposterior plane, there were no significant between-movement differences for forwards ($p \geq 0.80$) or backwards ($p \geq 0.37$) accelerations. However, analysis of mediolateral motion identified that the échappé (0.55 ± 0.28 g) had significantly lower right accelerations compared with the jeté step (0.94 ± 0.34 g, mean difference = 0.39 g; CI for difference: 0.69-0.09, $p = 0.01$, $d = 0.53$), sissonne (0.72 ± 0.22 g, mean difference = 0.17 g; CI for difference: 0.32-0.01, $p = 0.03$, $d = 0.32$) and jeté ET (1.05 ± 0.59 g, mean difference = 0.50 g; CI for difference: 0.93-0.07, $p = 0.02$, $d = 0.48$) respectively. There were no significant between-movement differences ($p \geq 0.11$) for left accelerations. Finally, in the vertical plane, upwards accelerations during the jeté step (3.79 ± 0.80 g) were significantly greater compared with the sissonne (2.74 ± 0.56 g, mean difference = 1.05 g; CI for difference: 0.06-2.04, $p = 0.03$, $d = 0.61$), and during the jeté ET (3.29 ± 0.69 g) compared with the sissonne (mean difference = 0.56 g; CI for difference: 0.02-1.10, $p = 0.04$, $d = 0.40$). Analysis of downwards accelerations highlighted significantly greater values during the jeté (0.76 ± 0.35 g) compared with the temps levé (0.44 ± 0.22 g, mean difference = 0.32 g; CI for difference: 0.14-0.62, $p = 0.04$, $d = 0.48$). The movement x plane x direction x limb interaction was not significant ($F = 9.12$, $p = 0.54$). Resultant symmetry indexes from the peak acceleration values in each plane are displayed in the Table 7.3.

Table 7.3. The influence of movement task and limb on symmetry indexes for each plane derived from the peak acceleration values. The data presented are mean values.

Variable	Limb	Ballet Jump Variant						
		Jeté	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps levé**	**Jeté ET**
ACC_{AP}	Dom	0.32	0.27	0.32	0.26	0.29	0.28	0.31
	Non-Dom	0.28	0.29	0.35	0.26	0.29	0.25	0.30
ACC_{ML}	Dom	1.20	1.17	0.88	1.22	1.09	1.09	1.24
	Non-Dom	0.96	1.17	1.10	0.91	0.99	1.25	1.31
ACC_{V}	Dom	4.77	4.85	5.67	5.27	4.68	6.43	6.35
	Non-Dom	4.04	3.62	4.65	6.20	5.69	7.22	6.51

Dom, Dominant; Non-Dom, Non-Dominant; ACC_{AP}, anteroposterior accelerations; ACC_{ML}, mediolateral accelerations, ACC_{V}, vertical accelerations.

7.4 Discussion

The primary aim of the current study was to comprehensively quantify and evaluate parameters of acceleration and load during the same ballet-specific movement battery used in studies 1 and 2. State of the art motion tracking via force plate and automated 3D motion capture may not be suitable for ballet training and competition. Hence, innovation in the use of alternative technologies may offer the potential for rigorous assessments of ballet movement in field-based settings.

The first key observation was the significant main effect for movement on planar and accumulated total body load across the jump landing test battery. Within existing literature, there is no other data available on accelerometer-derived load responses to various ballet manoeuvres. Hence, this finding is difficult to contextualise and cannot be generalised beyond the current investigation. However, the between-task variability in PlayerLoad^{TM} reflects the findings of research in a different sport modality (Barreira et al., 2017), with accelerometer analysis distinguishing between jogging, side cutting, striding, and sprinting tasks. Accelerometer responses in the current study reflected the hierarchical ordering of movement characteristics inherent to each jump landing task. For example, PL_{TOTAL} elicited during the jeté step manoeuvre was 31.85%-118.18% greater than the jeté and sissonne techniques respectively, but there was no difference in PL_{TOTAL} between the jeté step and jeté ET. The greater magnitude of accumulated load during the jeté and jeté ET reflects the kinetic demand highlighted in the data obtained during study 1. Further, the biomechanical hierarchy of task response within the movement battery is emphasised in acknowledgement of the lower PL_{TOTAL} responses during the échappé, sissonne and sissonne PDB, which again mirrors the kinetic data of study 1. With Newton's law of acceleration (2^{nd} law) stating that resultant force is a product of mass and velocity, the greater accelerations during the jeté step and jeté ET are expected. The magnitude of load responses may also be influenced by the kinematic profile specific to each ballet task. Performing the jeté ET requires movements in all three planes, whereas the échappé, sissonne and sissonne PDB predominantly have uni-axial kinematics in either vertical or mediolateral displacement. The lack of comparable studies in ballet-specific movements limits a critical discussion of these observations, underlining the need for further research in this domain.

The battery of tasks used in the current study was designed to differentiate between hold and transitional landings, and techniques performed uni- or bilaterally. The jeté step manoeuvre required participants to execute a jeté but to maintain anterior momentum via a landing transition rather than a hold. Intuitively, and as demonstrated in planar analysis, the transition would produce increased anteroposterior accelerations and subsequent load. However, there were greater responses in the vertical component also, which may be symptomatic of a greater commitment to the aerial aspect of the manoeuvre. This finding has improved performance and clinical interpretation when placed in context with the kinetic findings presented earlier (Chapter 4) within the current thesis. There were no significant differences in peak vertical GRFs between the jeté and jeté step, but the accelerometer data of this study shows greater vertical accelerations and load in the jeté step which may represent increased jump height. Including a transitional landing element may encourage a more explosive execution of the jeté to appease performance aesthetics without amplifying impact forces, and, the ensuing injury risk. It must be noted that jump height was not measured in the current study, and thus, this observation remains speculative and should be treated with caution. Nevertheless, in a training scenario whereby ballet dancers perform many vigorous jump landing tasks for prolonged periods (Twitchett, Koutedakis and Wyon, 2009), distinguishing the mechanical load responses between techniques may valuably inform workload and task prescription towards managing injury risk. Accelerometers are sensitive to the planar variances and demands differentiating tasks, and may enhance current ballet performance monitoring strategies through quantifying the loads imposed by select ballet movements.

Individual planar contributions to PL_{TOTAL} were also quantified to provide a greater insight into the kinematic responses to the discrete ballet manoeuvres. The variation in planar demand between movements was reflected in a significant main effect for uni-axial contribution. The greater input of PL_V to PL_{TOTAL} is perhaps reflective of the greater emphasis on aerial displacement during ballet-specific jump landing tasks, in acknowledgement of the artistic component of performance. The significantly greater PL_{AP} contribution compared with PL_{ML} for all movements except the jeté ET may reflect the movement direction characterising each jump. Further, it may represent a particular strategy used in the cessation of momentum and in providing whole-body stability during landing. This effect seemed to be more demonstrable during the échappé technique, in which greater PL_{AP} but smaller PL_{ML} contributions were evident compared with other tasks. The échappé jump in the current study initiated with the feet placed in turnout, required a vertical jump, and terminated with a landing in the same foot configuration. During turnout, the foot pronates, and the ankle complex everts to provide stability (Carter, Bryant and Hopper, 2019), which is particularly important given the everted foot configuration during the échappé landing as evidence by the kinematic findings of study 1. The greater relative PL_{AP} during the échappé compared with other tasks appears to reflect the compensatory mechanism for compromised stability resulting from turnout.

Although uni-axial load analysis provides a greater level of detail when quantifying the loads imposed by sports locomotion, the sign convention issue in squaring each vector component limits resultant interpretation. From a force plate analysis perspective, this step is akin to summing the positive and negative force signals to produce a net GRF. However, the mediolateral force vector as an example represents inversion and eversion which must be considered in isolation, and these principles should translate to accelerometer-based analyses. The uni-axial load equation is configured to describe motion as forwards, right and upwards, a movement description which almost never stands true in sport. In the current study, within-planar accelerations were quantified to delineate the magnitude of movement in a particular direction. The novel symmetry index presented within, highlighted a ~3:1 imbalance in the anteroposterior plane towards greater backwards motion. This observation may indeed reflect posterior trunk lean typical during gait to govern stability when ceasing forward momentum (Powers, 2010). There was no significant difference in mediolateral movement (~1:1) direction which further strengthens the argument that ballet dancers stabilise the body predominantly in the anteroposterior plane when landing from a jump. In the vertical plane, upwards accelerations outweighed downwards by ~5:1 reflecting the magnitude of the aerial aspect comprising each jump landing task. A significant between-movement difference was identified in downwards accelerations, with the jeté observing significantly greater downwards motion compared with the temps levé. Though not significantly different to other movements, a similar trend towards greater downwards acceleration was observed during the jeté step which is possibly indicative of a more pliable landing strategy to attenuate the high impact forces of this technique as demonstrated during study 1. The lower values identified for the échappé and sissonne may be due to the lower impact forces highlighted by kinetic analyses, or representative of a stiffer landing technique. Very few studies have referenced nor considered the movement direction relating to each individual vector. Consequently, a thorough understanding on the locomotor and subsequent load profiles of selected tasks is prohibited without this level of analysis. The current findings shed light on the importance of considering within-planar accelerations in comprehensive assessments of movement using accelerometry.

To be consistent with the previous experimental chapters of the current thesis, and to acknowledge the asymmetric movement profile of ballet (Liederbach et al., 2006), the influence of laterality on accelerometer-derived load responses was considered in analyses. A second key finding highlighted bilateral symmetry in acceleration and load responses across the movement battery, reflecting the data observations from the previous experimental chapters of the thesis. There was no significant bilateral difference in planar indices of load, and thus, a lack of asymmetry in the uni-axial contribution to PL_{TOTAL}. PlayerLoad^{TM} is calculated using the magnitudes of acceleration in the three movement vectors (Waldron et al., 2011). The evident symmetry in peak accelerations therefore, underpins the symmetry in accelerometer-derived load responses to ballet-specific jump landing tasks. Practitioners may therefore infer that ballet dancers adopt the same kinematic landing strategy, irrespective of the limb used to execute discrete tasks. The lack of asymmetry in accelerometer loading in this ballet populations may be attributed to early onset of ballet training, with emphasis placed on lower limb control and trunk stability when performing the movements comprising a routine (Bronner and Ojofeitimi, 2006; Mertz and Docherty, 2012).

Although there was some control over the distance each participant initiated the discrete techniques, approach velocity and resultant jump height was self-determined. Despite the homogeneity in dancer skill level, and that choreographed routines dictate a specific tempo, there may be some disparity in the velocity of movement and jump height achieved between dancer which may influence the resultant kinetic and kinematic response. Acceleration and load responses to the multi-task, ballet-specific movement battery were quantified using the conventional mid-scapula device placement which serves as a potential limitation to the current study. The single unit may not be the most appropriate method given that the majority of movements have a multi-segmental construct (Nedergaard et al., 2017). In addition, anatomical consideration when selecting unit location has been shown to influence resultant loads and the interpretation of movement (Barrett et al., 2016; Greig and Nagy, 2017; Brogden et al., 2018), and therefore, inclusion of additional units in the current study may have yielded different findings. The absence of similar analyses in a ballet context prohibits a comparison of the findings with those of other studies, and therefore, the current observations should be treated with this in mind. Nevertheless, the experimental approach undertaken in this investigation highlights innovation and rigour in the assessment of ballet movement. The portability of accelerometry improves ecological validity, and the comprehensive tri-planar evaluation of mechanical load offers greater biomechanical application in settings beyond the laboratory.

7.5 Conclusion

The current study is the first to present an accelerometer-derived load profile of tasks forming a typical ballet performance repertoire. Further, it represents one of very few examples to consider the direction of movement within each planar vector, highlighting innovation and rigour in load assessment. The accelerometers response in PL_{TOTAL} and planar load with corresponding relative contributions, were sensitive to the kinematic characteristics specific to each task, and, the hierarchy in movement demand mirrored the kinetic and kinematic observations of study 1. Analysis of peak accelerations within each movement plane showed greater upwards accelerations owing to the explosive aerial component of jump landing tasks, and, greater posterior accelerations representing trunk movement when ceasing forward momentum. Within-planar accelerations analysis provides a greater understanding on the strategy used to perform discrete ballet movements. The symmetry in accelerometer-derived load responses between limbs simulates the findings from the previous experimental chapters, and may suggest that current ballet training practices facilitate bilateral development in ballet movement execution. Accelerometers have good portability compared with 3D motion capture and force plate analysis, and mirror the mechanical responses of traditional analysis tools used in the previous studies. This utility may support and enhance field-based evaluations, and subsequent investigations into associations between accelerometry- and laboratory-based analyses could strengthen this potential.

Chapter 8. Study five: Quantifying the Association between Field- and Laboratory-based Measures of Loading during Ballet-specific Movements.

8.1 Introduction

The previous experimental chapter quantified accelerometer-derived load responses to the same battery of movements used in studies 1 and 2. The findings showed that accelerometry was sensitive to the ballet-specific movements in support of the kinetic and kinematic observations, but importantly, that the ordering of mechanical demand was the same between studies. This pattern in biomechanical response demonstrates an association between the tools of analysis, but accelerometry offers portability over laboratory-based methods and may serve as useful means of conducting biomechanical assessments in traditional performance environments. Whilst qualifying associations highlights the rigour in accelerometer-based analyses, quantifying a correlation between data collection methods is the next progression towards valid biomechanical assessments in field-based designs. In recent times, there has been a surge in the application of microtechnology as a measurement tool in sport (Raper et al., 2018), with ever-increasing research investigating the relationship between accelerometer-derived and traditional methods of load assessment. One of the most common relationships to have been explored in relevant studies is that between vertical accelerations and the vertical component from GRF analysis. This correlation has been quantified in a number of gait activities including walking (Shahabpoor and Pavic, 2018), running (Edwards et al., 2019) and jumping (Simons and Bradshaw, 2015; Setuain et al., 2016). There is scope to adopt similar methods in ballet-specific motion which may improve current monitoring practices to better understand the mechanical loads experienced by dancers in training and performance.

Execution of the manoeuvres comprising a typical ballet routine generate GRFs that transmit a transient shockwave through the kinetic chain (van der Worp, Vrielink and Bredeweg, 2016). The magnitudes of GRFs, specifically the vertical component, have been associated with lower limb injuries (Hreljac, 2004), and, ballet dancers with lower limb pathology have demonstrated higher GRFs during landing tasks compared with uninjured counterparts (Fietzer, Chang and Kulig, 2012; Peng et al., 2015). The gold standard technique for quantifying external load during human movement is force plate analysis, and several investigations have been conducted in ballet populations (Kulig, Fietzer and Popovich, 2011; Mertz and Docherty, 2012, Jarvis and Kulig, 2016). However, due to their complex operational process and their configuration within a laboratory environment, the constraints of force platform analysis limits the capacity to replicate true ballet motion (Roell et al., 2019). Hence, existing methodologies have reduced specificity and ecological validity in research design, and alternative methods to access data more readily in performance settings are desirable.

Intuitively, accelerometers may be used for indirect estimations of GRFs based on the relationship between force and acceleration according to Newton's second law of motion (F = m·a) (Verheul et al., 2020). Accelerometer devices are worn on the posterior aspect of the upper trunk and record corresponding segmental accelerations. In previous studies, there is conflicting evidence regarding the utility of trunk-mounted accelerometry to quantify external load in running (Wundersitz et al., 2013; Nedergaard et al., 2017; Edwards et al., 2019), but strong, positive vertical acceleration-vertical GRF magnitude correlations have been reported in jump landing tasks (Simons and Bradshaw, 2016). The application of accelerometry therefore, offers promise in the assessment external load, and may alleviate some of the limitations associated with laboratory testing environments. In addition, the strong acceleration-GRF correlations shown in jump landing tasks suits the movement profile of ballet, and accelerometers may serve as attractive alternative to classic force platform analysis in ballet.

Although strong, positive correlations have been identified in other clinical jump landing trials (Simons and Bradshaw, 2016), relatively little is known about the accelerations-GRF relationship in ballet-specific movements. Therefore, the aim of the current study was to determine the association between trunk-mounted vertical accelerations and peak vertical GRFs and loading rates during a range of ballet-specific jump landing tasks.

Experimental research question:

1) Does trunk-mounted accelerometry associate with kinetic measures of load, and if so, what is the relative strength of the association?

8.2 Methodology

Participants

The current study presents a secondary analysis of the GRF data obtained during study 1, and the accelerometer responses in study 4. Therefore, data from the same cohort of 12 female dancers was used.

Data Processing

The integrated accelerometer (Kionix KX94, Kionix, Ithaca, New York, USA) of an athlete tracking technology was used to quantify resultant accelerations, with data sampling at 100 Hz. Unfortunately, the authors were unable to acquire information pertaining to the in-built filtering methods, and therefore, resultant accelerometer data was not further filtered after being exported. Post-processing was conducted in accompanying software (Catapult Sprint), and peak vertical accelerations (vACC) (g) were identified for each landing with corresponding values representing impact accelerations. Manual synchronisation of the data sets was achieved using a pre-determined event, specifically a series of 10 vertical jumps prior to initiation of the experimental trials. The time of data capture in QTM was compared the corresponding time displayed within each accelerometer device, and further synchronisation was ensured by comparing number of acceleration peaks within each movement with the number of peak magnitudes in the vertical GRF signal. Thus, the author was confident that each acceleration peak was aligned with the correct landing repetition, for each ballet-specific jump landing task.

Statistical Analysis

All data were analysed using a statistical software package (SPSS V25.0, IMB, Armonk, New York, USA). Data normality tests were conducted to assess the dispersion of data, and with the normality assumptions satisfied, the relationship between vACC and vertical ground reaction force (vGRF), and, vACC and vertical mean loading rate (vMLR) was investigated using the Pearson's correlation statistic. Correlation coefficients (r) were calculated to determine the strength of the relationships, with values of 0.10-0.29 representing a weak relationship, 0.30-0.49 moderate, 0.50-0.69 strong, and ≥ 0.70 very strong (Hopkins et al., 2009). Statistical significance of the correlation coefficients was also examined, with the threshold set at $p < 0.05$.

8.3 Results

Descriptive statistics for each biomechanical output measure are displayed in Table 8.1.

Table 8.1. Data for the identified biomechanical parameters for each ballet-specific jump landing task. Corresponding values are mean ± σ.

	Ballet Manoeuvre						
Variable	**Jeté**	**Jeté step**	**Échappé**	**Sissonne**	**Sissonne PDB**	**Temps leve**	**Jeté ET**
vACC (g)	3.42 ± 0.93	3.18 ± 0.85	2.69 ± 0.64	2.56 ± 0.42	2.52 ± 0.62	2.88 ± 0.44	3.26 ± 0.58
vGRF ($N \cdot kg^{-1}$)	2.97 ± 0.60	2.99 ± 0.83	1.90 ± 0.50	2.65 ± 0.60	2.49 ± 0.63	3.06 ± 0.70	3.24 ± 0.69
vMLR ($BW \cdot s^{-1}$)	46.60 ± 16.87	52.13 ± 26.90	29.36 ± 12.15	30.14 ± 10.59	29.46 11.27	32.28 ± 13.16	40.97 ± 14.57

vACC, vertical accelerations; g, accelerations; $_V$GRF, vertical ground reaction force; $N \cdot kg^{-1}$, Newtons per kilogram; $_V$MLR, vertical mean loading rate, $BW \cdot s^{-1}$, body weights per second.

Scatterplots portraying the linear relationship between vACC and $_V$GRF are displayed in Figure 8.1. The Pearson's correlation test revealed significant very strong, positive relationships between vACC and vGRF for the jeté ($r = 0.81$, $p < 0.01$), and temps leve ($r = 0.77$, $p < 0.01$), and significant strong, positive correlations were identified for the jeté step ($r = 0.65$, $p = 0.02$), échappé ($r = 0.59$, $p = 0.04$), sissonne ($r = 0.69$, $p = 0.01$), sissonne PDB ($r = 0.59$, $p = 0.04$), and jeté ET ($r = 0.63$, $p = 0.03$). Scatterplots of the vACC and vGRF for each manoeuvre are presented in Figure 8.1.

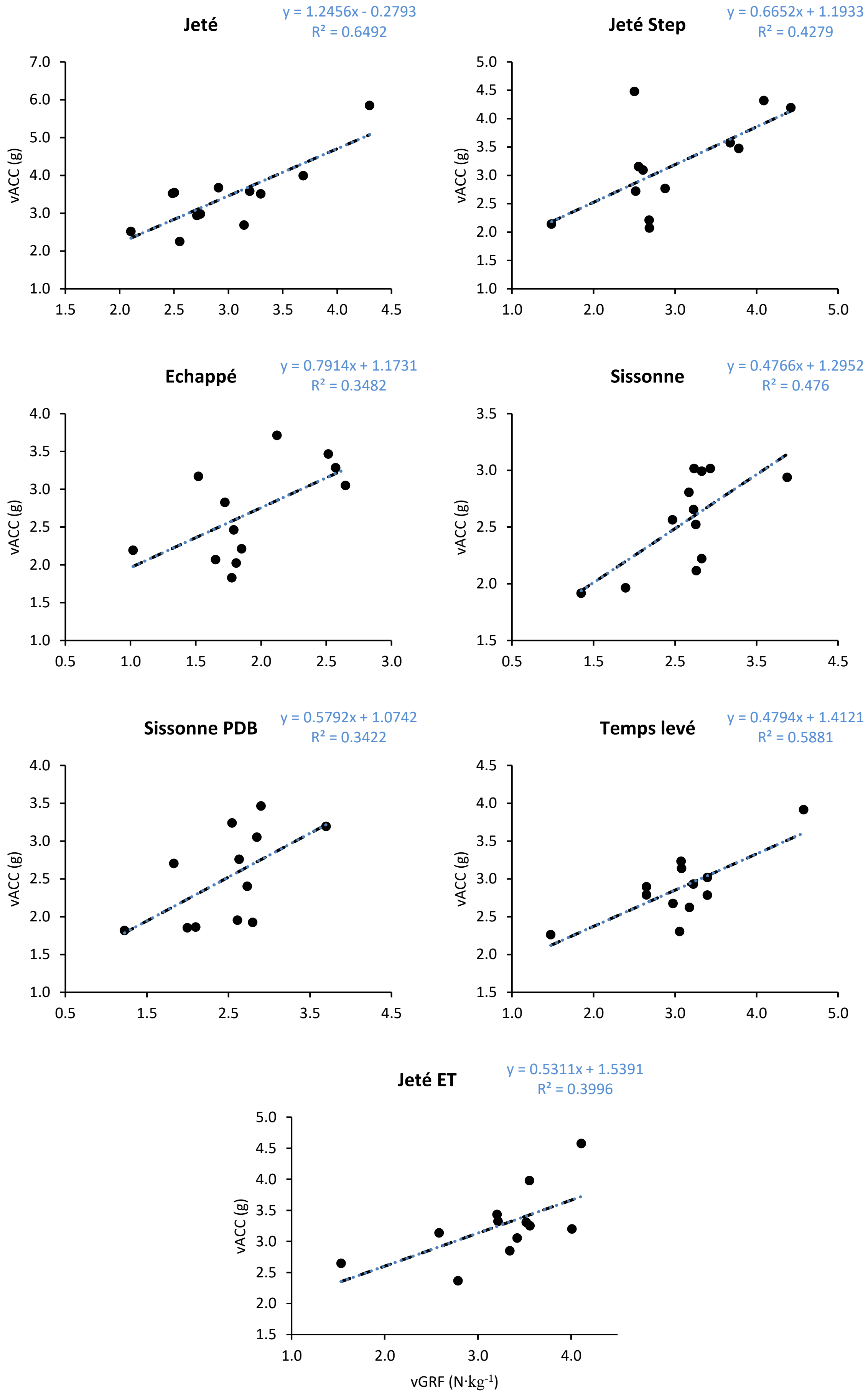

Figure 8.1. Scatterplots of the impact accelerations and peak vertical ground reaction forces.

Figure 8.2 presents scatterplots showing the linear relationship between vACC and vMLR. Pearson's correlation analyses identified significant very strong, positive relationships between vACC and vMLR for the jeté ($r = 0.82$, $p < 0.01$), temps levé ($r = 0.75$, $p = 0.01$) and jeté ET ($r = 0.84$, $p < 0.01$). Further, there was a significant strong, positive correlation for the sissonne PDB ($r = 0.59$, $p = 0.04$). However, no significant relationship was revealed for the jeté step ($r = 0.54$, $p = 0.07$), échappé ($r = 0.56$, $p = 0.06$), or the sissonne ($r = 0.51$, $p = 0.09$). Scatterplots of vACC and vMLR for each participant and for each task are displayed in Figure 8.2.

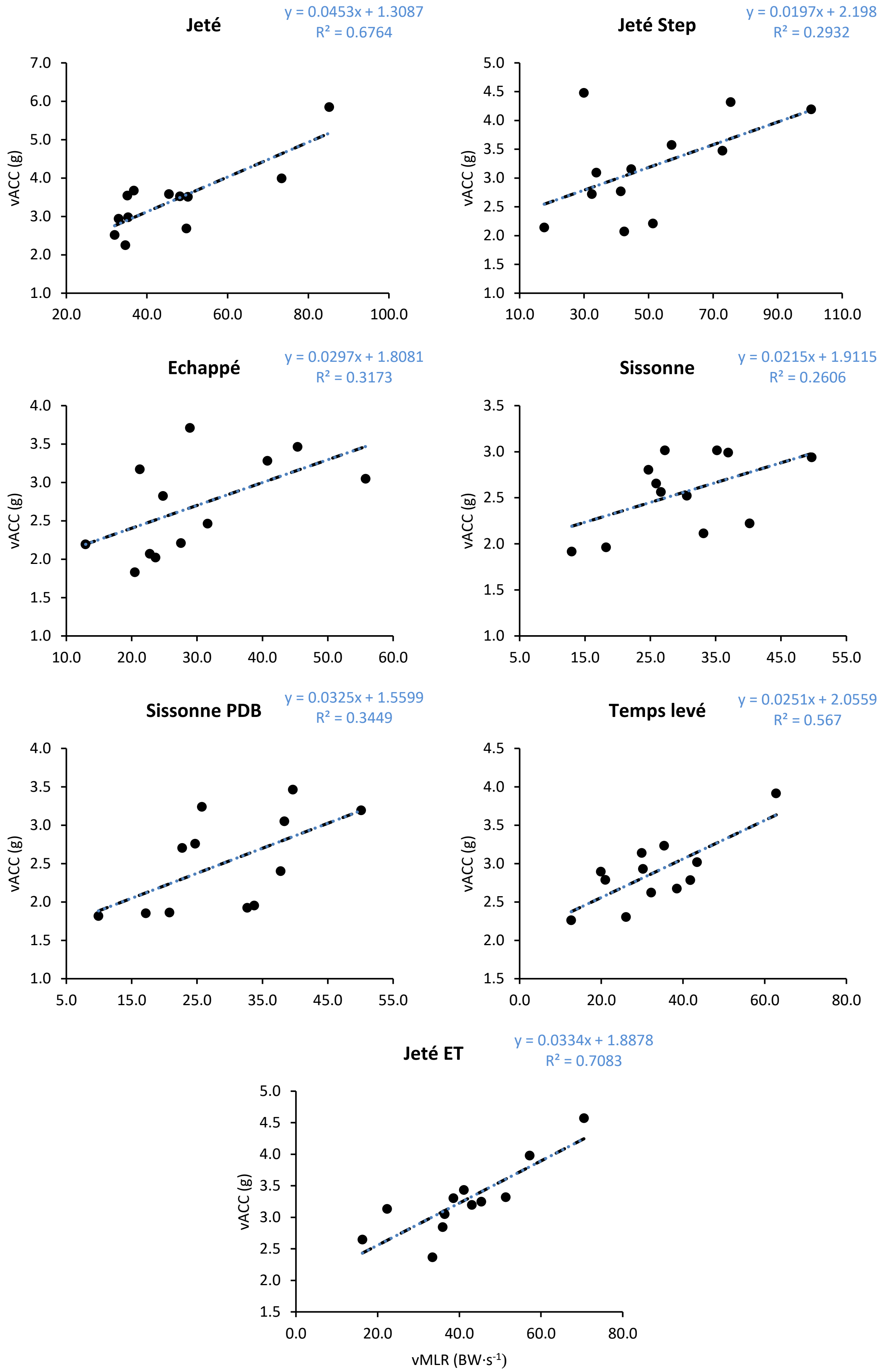

Figure 8.2. Scatterplots of the impact accelerations and vertical mean loading rates.

8.4 Discussion

The aim of the current study was to investigate the relationship between trunk-mounted accelerometry and the impact magnitudes and loading rates of the vertical ground reaction force vector during a seven-task, ballet-specific movement battery. Analyses highlighted significant, very strong to strong positive correlations between impact accelerations and peak vertical ground reaction forces for all manoeuvres. Moreover, significant, very strong to strong, positive relationships were identified between impact accelerations and mean vertical loading rates.

The strength of the accelerometer-vGRF relationship between tasks of the current study ranged from strong for the sissonne PDB ($r = 0.59$) to very strong for the jeté ET ($r = 0.81$). These findings are consistent with those reported in a similar study (Almonroeder et al., 2019), however the acceleration-GRF correlation was demonstrated in a broad range of ballet manoeuvres, compared with the single changement de pied technique used in the aforementioned study. The disparity in coefficient values were generally proportionate to the degree of vertical displacement specific to each task. For example, the strongest correlations ($r \geq 0.77$) were observed for the jeté and temps levé trials which have a greater requirement for vertical movement during execution. The manoeuvres that required a lateral displacement of the body (sissonne), incorporated a rotational element (jeté ET), or indeed, contained a transitional landing component (jeté step, sissonne PDB) observed lower correlation coefficients. However, it is important to note that only the vertical acceleration vector was examined in the current study which may have influenced the discrepancies in the strength relationships observed between tasks. Hence, consideration of net accelerations may have yielded different results, and interpretation of the findings beyond the specific methodological design should be treated with this in mind. Nevertheless, the strong, positive relationships between vertical accelerations and the corresponding vector of GRF for all tasks, suggests that accelerometry offers a promising method of assessing external forces during ballet-specific movements and may enhance current load monitoring strategies.

The current investigation relates most closely with the research by Almonroeder and colleagues (2019), given the ballet specificity consideration and similarity in task selection, although the correlation coefficients are comparatively lower ($r \geq 0.59$ vs $r \geq 0.95$). Discrepancies in the relationship strength between the studies may be partially explained by the tasks incorporated. For example, the sissonne and sissonne PDB techniques used in the current study have a markedly different kinematic profile to that of the changement de pied. The sissonne and sissonne PDB requires a lateral displacement, whereas the changement de pied requires a dancer to assume 5th position (turnout, with the toes placed behind/in front of the heel of the contralateral foot) and jump vertically whilst changing the position of the feet before landing (Almonroeder et al., 2019). The most comparable manoeuvre to the changement de pied from the current study was the temps leve (a vertical hop) which demonstrated a very strong, positive relationship (r = 0.77). The strength of the association between the vertical accelerations and the vertical GRF component may be influenced by the placement of the accelerometer device at C7. The temps levé, which observed one of the strongest coefficients, is characterised by a vertical hop with minimal displacement of the lower limbs about the trunk. The movements with the weaker strength correlations (échappé, sissonne, sissonne PDB) require a split formation of the legs, and a lateral displacement of the body. Hence, locating the accelerometer device at the conventional C7 position may limit the interpretation of movement, and may partially explain the variability in coefficients between tasks. A further suggestion regarding the varying association strengths may be linked to accelerometer data sampling rates, in that the current study used a lower frequency (100 Hz) which may not be optimal when quantifying accelerations during jump landing tasks (Ziebart et al., 2017).

Many investigations have quantified the relationship between impact accelerations and ground reaction force magnitudes, with a variety of device attachment locations and locomotor tasks evident in pertinent studies (Wundersitz et al., 2015; Simons and Bradshaw, 2016; Setuain et al., 2016; Nedergaard et al., 2017; Edwards et al., 2019). Attachment location of the unit when investigating the application of accelerometery to estimate ground reaction forces is a contentious issue. It has been suggested that a distal location in the kinetic chain (i.e. shank) provides the most valid data (Lafortune, Lake and Henning, 1996; Edwards et al., 2019), whilst a bony anatomical landmark is deemed more preferable for reliability to mitigate soft tissue artefact (Yang and Hsu, 2010). This configuration contrasts with the conventional recommendation of thoracic placement within the elasticated pouch of a manufacturer provided vest. Previous studies in running have shown that the trunk-mounted device method provides a poor estimation or resultant ground reaction forces (Nedergaard et al., 2017; Edwards et al., 2019), with the authors postulating that accelerations are augmented via a device ‘whipping’ mechanism. Linear running is characterised by anteroposterior motion which likely far exceeds the magnitudes of vertical displacement. That only the vertical acceleration vector was incorporated into analyses by Edwards and colleagues, (2019), may explain the lack of an acceleration-ground reaction force relationship. This argument is strengthened by the strong positive associations demonstrated in research using jump landing tasks (Simons and Bradshaw, 2016), and indeed, the findings from the current investigation. These observations suggest that the utility of trunk-mounted accelerometers to indirectly assess ground reaction forces may indeed be task dependent, although further research is required to substantiate the suggestion. A trunk-mounted strategy was adopted for the current study, representing a location with minimal impact on the artistic and visual aesthetics of ballet performance. Securing the devices to other anatomical locations is typically achieved using modified strategies such as semi-elastic straps (Ziebart et al., 2017) and tape (Simons and Bradshaw, 2016; Greig and Nagy, 2017), both of which may not adequately stabilise the device. Moreover, these non-rigid attachment methods may not be appropriate for the explosive multi-planar jump landing tasks inherent to ballet, or for prolonged periods of performance monitoring. To that end, developments in material design and harnesses to facilitate multi-site acceleration recordings may improve the validity and reliability of current methodologies.

Progressing from laboratory- to field-based approaches in biomechanics presents a challenge, given the difficulty in quantifying biomechanical loads (Verheul et al., 2020). This is perhaps exacerbated by a lack of available methods to validly obtain biomechanics information in a natural training and/or competition environments such as a ballet studio. Observations from the current investigation propose that accelerometers may serve as a valuable biomechanical tool for determining external loads. The findings suggest that peak vertical accelerations provide a valid indication of the forces produced during ballet-specific jump landing tasks. The added biomechanical dimension to motion tracking offered by accelerometry may be used to monitor volume and intensity during congested training periods. High injury rates have been established in ballet, with those attributed to overuse representing a primary concern (Smith et al., 2015). Accelerometery-based analyses may therefore provide important aetiology information regarding the occurrence of injury, whilst informing the strategies used in managing risk. Despite the current study highlighting the potential for wearable accelerometers to indirectly assess GRFs in ballet dancers during a variety of jump landing tasks, there are limitations that need to be acknowledged. First, only the relationship between vertical accelerations and kinetic response were considered in the correlation analyses. Data relating to the capacity of accelerometry to validly estimate joint kinematics would provide greater insight into movement execution. The combination of kinematic and kinetic information would strengthen the assessment of the external load imposed by ballet, thereby facilitating the transition from the laboratory to training and competition environments. Second, there is currently a lack of consensus regarding the optimal location of the accelerometer device for indirectly assessing GRFs during sport movements. The addition of another unit(s) at other locations in the current study may have provided a more rigorous analysis and furthered the understanding of the acceleration-GRF relationship.

8.5 Conclusion

This study is the first to investigate the relationship between vertical accelerations recorded at the upper trunk and vertical ground reaction forces during a range of ballet-specific jump landing tasks. The respective correlations were highlighted to be strong to very strong (r = 0.59-0.84) for all movements, with the variability in coefficients reflecting the planar characteristics specific to each task. The correlation coefficients of less than 1 suggests that other factors beyond the vertical acceleration component may influence the strength of the accelerometry-GRF relationship, thereby prompting the need for further investigation. Nevertheless, accelerometers may provide an indication of the external loads imposed by performing ballet-specific movements. Accelerometery provides a portable experimental data collection tool that enables a multi-planar analysis of movement, offering scope to develop monitoring strategies and the application of biomechanics in real performance settings. Monitoring the volume and intensity of ballet-specific training and competition using accelerometry may have clinical implications towards understanding the occurrence of injury.

Chapter 9. Study six: Within- and Between-day Load Responses to Ballet Training and Performance

9.1 Introduction

To this point, the studies comprising this thesis have demonstrated consistency in the observation of a task-specific response across a multi-modal biomechanical evaluation of ballet-specific movement. Further, the bilateral symmetry evident in this cohort of female dancers has also been highlighted across the range of data collection tools. The potential for field-based biomechanical analyses of ballet was demonstrated in the previous experimental chapter, and in this final investigation, the biomechanical demands of ballet are considered in a more ecologically valid context.

For ballet dancers, extensive training regimens initiate at an early age, and technical development thereafter requires many hours dedicated towards training and competition (Wyon and Koutedakis, 2013). This unrelenting schedule is unconducive to rest, and augments the possibility of fatigue-induced ‘burnout’ which may contribute to overuse injury (Murgia, 2013). Dancers may not complete a sufficient rest period between bouts, a notion strengthen by previous research on the daily activity profiles of ballet dancers highlighting that around 90% take less than 60 minutes rest, and 33% take less than 20 minutes rest (Twitchett et al., 2010). Other study findings have shown more injuries to occur in the latter stages of a performance, and towards the end of a season, thereby implicating fatigue with injury occurrence (Liederbach, Dilgen and Rose, 2008). In other sports, strategies to reduce the number of injuries attributed to overuse injury include restrictions on training and competition workloads (Schaefer et al., 2018). Currently, there is no governing body legislation advocating limits on exposure to ballet. Yet, ballet typically demands high training volumes, and dancers may be required to perform choreographed routines across consecutive days of a week (Wyon and Koutedakis, 2013). Avoiding the perilous combination of excessive workloads and inadequate rest is crucial in ballet training prescription, helping to mitigate performance decrements and an increased injury risk (Meeusen et al., 2013, Soligaard et al., 2016; Schwellnus et al., 2016). To that end, efforts towards developing a greater understanding on the cumulative mechanical effects of ballet training are worthwhile.

As evidenced within the literature review section of the thesis (Chapter 2.3), existing biomechanical analyses in ballet typically comprise a laboratory-based 3D kinematic and kinetic evaluation of movement. Whilst these environments offer high control and enable sophisticated analyses, previous designs have either incorporated a clinically-focused drop landing task (Orishimo et al., 2009, Liederbach et al., 2014), or a solitary ballet-specific manoeuvre (Kulig, Fietzer and Popovich, 2012, Peng et al., 2015, Jarvis and Kulig, 2016). Improved specificity in research methodology was demonstrated in study 1 via the development of a battery of ballet-specific jump landing tasks of varying planar demand. However, study 1 quantified the acute responses to discrete ballet techniques given the constraints of laboratory-based testing, and highlighted the need for alternative methods to quantify load in more ecologically valid settings. As shown in studies 4 and 5, accelerometry is sensitive to ballet task characteristics and planar demand, and provides an indication of the external loads imposed during movement. Accelerometry has been used extensively in team invasion sports (Boyd, Ball and Aughey, 2013; Dalen et al., 2016), providing a means to identify sub-optimal training loads that have been associated with an increased risk of injury (Colby et al., 2014; Bowen et al., 2017). Consequently, there is scope for similar investigations in ballet, and monitoring dancer capacity towards high training loads using accelerometry may provide key information in injury aetiology.

Along with the tools used in biomechanical analyses, the call for greater ecological validity and specificity in research design has resulted in the use of sport-specific experimental protocols such as the 90-minute soccer aerobic fitness test ($SAFT_{90}$) in soccer research (Small et al., 2009), and more pertinently, the dance aerobic fitness test (DAFT) (Wyon et al., 2003). In a recent study, Brogden and Colleagues, (2018) investigated accelerometer-derived load responses to the DAFT protocol. Their findings demonstrated that mechanical load increased proportionately to exercise duration and intensity. Further, anatomical consideration of unit placement showed a level of sensitivity towards greater resultant loads in the lower limb compared with the recommended upper trunk location. Despite these findings highlighting scope for quantifying the biomechanical responses to dance performance in authentic settings, load responses were investigated during a single trial. The methodological design therefore negates the repeated performance constructs of typical ballet training programmes, and prohibits an understanding on the cumulative effects of dance. In addition, although Brogden et al., (2018) quantified accelerometer responses in the lower limb to reflect the incidence of injury, this comprised a unilateral approach, yet, ballet has an asymmetric movement profile and thus warrants a bilateral lower limb analysis.

In light of these observations, the aims of the current investigation were two-fold. First, to examine within-and between-day accelerometer responses to a choreographed ballet routine comprising multiple stages of progressive intensity. Second, to quantify the sensitivity of accelerometry to different anatomical sites, with emphasis on lower limb laterality. Potential findings may develop knowledge on the workload tolerances of ballet dancers whilst supporting the use of contemporary athlete monitoring practices.

Experimental research questions:

1) Is accelerometry sensitive to a ballet-specific choreographed routine comprising multiple stages of progressive intensity?
2) Does a consideration of anatomical-specific loading influence mechanical response and the resultant interpretation of performance rigours?
3) Do female ballets dancers demonstrate altered profiles in mechanical load to consecutive choregraphed routines?

9.2 Methods

Participants

Data from pilot testing was used to conduct an *a prior* power analysis using G*Power software (v 3.1, Heinrich-Heine-Universität, Dusseldorf, Germany). Measures in accumulated PlayerLoad^{TM} revealed a medium (d = 0.54) effect size (Cohen, 1988), and with alpha set at $p < 0.05$ for significant differences with an observed power of 0.8, a minimum sample of 8 dancers was required to observe this effect. Following a random stratified sampling method, 10 female ballet dancers (Age: 23.20 ± 3.08 years; Height: 164.45 ± 5.46 centimetres; Body mass: 65.31 ± 8.47 kilograms) volunteered and provided written informed consent to take part in the study in respect of the Declaration of Helsinki. Eligibility for participation was subject to adherence with the inclusion criteria outlined in Chapter 3. Pre-exercise screening measures were completed in accordance with the protocols detailed in Chapter 3.

Experimental Design

All participants were required to attend the Dance Studio at Edge Hill University on three sperate occasions, comprising a familiarisation trial and two experimental trials. The familiarisation trial provided the opportunity for all dancers to complete each stage of a ballet-specific choreographed routine (see Table 9.1). The piece was taught and delivered by a qualified dance instructor with 15 years ballet performance experience, and 10 years ballet tutoring experience. The same ballet-specific routine was repeated for the two experimental trials, separated by 24 hours. All data collection sessions commenced at 10am to mitigate the effects of circadian rhythm on resultant performance. All participants were instructed to wear similar clothing and the same footwear to that worn during training and/or competition. Prior to the familiarisation and experimental trials, all participants completed a typical ballet class warm-up. The protocol consisted of barre and centre work prescribed by the qualified dance instructor, followed by self-selected dynamic stretching. All trials were conducted in the same dance studio, thereby simulating an authentic training and competition environment.

Ballet-specific Choreographed Routine

The choreographed dance routine below was designed to simulate a typical performance piece used in training and competition. The experimental protocol was choreographed in accordance with the principles of the DAFT protocol, in that it comprised multiple stages of progressed intensity. However, the routine contained some of the specific, multi-planar ballet techniques used in the methodological testing battery of studies 1, 2 and 4. A description of the movements comprising each stage of the piece is contained in Table 9.1.

Table 9.1. Technique descriptors and corresponding tempos for each stage of the ballet-specific choreograph.

Stage	Tempo ($b \cdot min^{-1}$)	Movement Description
1	70	Plié, relevé, sauté, changement, entrechat quatre, échappé, royale/changement battu
2	85	glissade, échappé, petit jeté, petit temps levé, temps de cuisse.
3	85	Failli, assemble (de coté devant), pas de chat, balloté, balloné, sissonne (double, ouvert, fermeé)
4	105	Grand temps levé, cabriole derriere, pirouette (en dehors), grand pas de chat, Grand assemble.
5	105	Grand jeté, posé (arrabesque/attitude derriere), grand assemble en tournant (en l'air), temps levé, developpé temps levé, grand sissonne ouvert turning, bournonville Jeté, coupé chassé en tournant, grand jeté en tournant, posé pirouette.

$b \cdot min^{-1}$, beats per minute

Each stage lasted 4 minutes and consisted of left and right limb execution, with an overall routine duration of 20 minutes. A one-minute period interspersed each stage. All elements of the experimental procedure were delivered by the same qualified dance instructor. The technical components and tempo of exercise were carefully controlled for in each stage, and chosen to demonstrate progressive intensity. The graded increment in intensity was achieved via manipulation of the discrete ballet manoeuvres, synonymous with petit and grand allegro.

Data Collection

Prior to the experimental trials, each participant was fitted with 3 GPS devices (Optimeye S5, Catapult Innovations, Melbourne, Australia). One unit was housed in a bespoke manufacturer provided neoprene vest and positioned at mid-scapula, approximating the 7th cervical vertebrae (see Figure 9.1). In consideration of the asymmetric kinetic movement patterns of ballet, and the sensitivity of accelerations to unit locations (Brogden et al., 2018), an additional unit was placed on the distal aspect of the dominant (DL) and non-dominant (NDL) leg. Under-wrap tape (Mueller Sports Medicine Incorporated, Wisconsin, USA) was used to secure the GPS device at a location 50% of the distance between the calcaneus and the posterior mid-point of the femoral epicondyles, approximating mid-gastrocnemius (see

Figure 9.1). The embedded accelerometer (Kionix KX94, Kionix, Ithaca, New York, USA) was used to record resultant accelerations, with data sampling at 100 Hz. Each participant completed a brief habituation period to prevent any compromised movement during the subsequent experimental protocol.

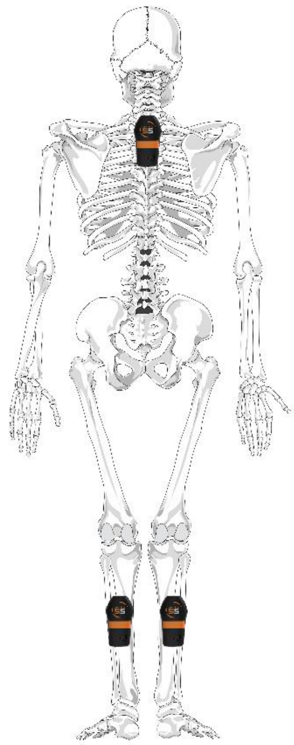

Figure 9.1. A representation of the approximate anatomical location and configuration of the Optimeye S5 GPS devices.

Physiological Responses

All participants wore heart rate (HR) monitors (HRM1G, Garmin, Kansas, USA) and corresponding watches (Forerunner 15, Garmin, Kansas, USA) to continuously record HR responses to the experimental trials. Borg's 6 to 20 point scale was used to quantify individual Ratings of Perceived Exertion (RPE).

Data Processing

Recorded accelerometer data was downloaded and exported to the corresponding GPS software (Catapult OpenField V1.17) for processing. A period for each stage of the choreographed routine was delineated to enable relevant accelerometer metrics to be extracted. Accumulated PlayerLoad^{TM} Total (PL_{Total}), defined as the sum of the PlayerLoad vector magnitudes in each movement plane (medial-lateral (PL_{ML}), anterior-posterior (PL_{AP}), and vertical (PL_V)), was calculated at C7 and the bilateral lower limb. Uni-axial planar contributions ($PL_{MP\%}$, $PL_{AP\%}$, $PL_{V\%}$) to total load were quantified by dividing each 1-dimnesional load value by total load and multiplying the resultant by 100.

Heart Rate (HR) and Rating of Perceived Exertion (RPE) were measured across the ballet-specific routine. Each participant wore a HR monitor (HRM1G, Garmin, Kansas, USA) and corresponding watch (Forerunner 15, Garmin, Kansas, USA), and individually referred to Borg's 6 to 20 point for RPE measures, with data recorded after each stage of the choreographed piece. Peak HR and RPE was selected for statistical analysis.

Statistical Analysis

All data parameters were analysed using a statistical software package (SPSS IBM Statistics V25.0, IBM, Armonk, New York, USA) and descriptive statistics are presented as mean ± σ. Histograms and the Shapiro-Wilk statistic were used to assess the dispersion of data. With the data normality assumption satisfied, a 2 x 5 x 3 x 3 (day x stage x unit position x uni-axial contribution) repeated measures ANOVA was used to measure within- and between-day differences in the selected accelerometer outcome measures during the ballet-specific dance routine. Separate 2 x 5 (day x stage) ANOVAs were conducted for the HR and RPE responses. Where appropriate, post-hoc comparisons using a Bonferroni correction factor were conducted to identify which variables were significantly different from each other. 95% Confidence Intervals (CI) are presented for any significant findings, along with Cohen's *d* effect sizes (small, 0.20-0.49; moderate, 0.50-0.79; large ≥ 0.80). Differences were deemed statistically significant at the $p < 0.05$ level.

9.3 Results

Choreographed Routine Load Responses – PL_{TOTAL}

Data on within-and between-day changes in PL_{TOTAL} across the ballet-specific dance routine is displayed in Figure 9.1. The ANOVA found no significant ($F = 2.08$, $p = 0.18$) difference in load responses between day 1 (118.44 ± 18.26 au; CI: 108.66-128.22 au) and day 2 (109.50 ± 15.78 au; CI: 101.92-117.08 au, $p = 0.18$), and no significant day *x* stage ($F = 0.96$, $p = 0.44$), day *x* unit position ($F = 0.84$, $p = 0.45$), or day *x* stage *x* unit position ($F = 0.43$, $p = 0.90$) interaction.

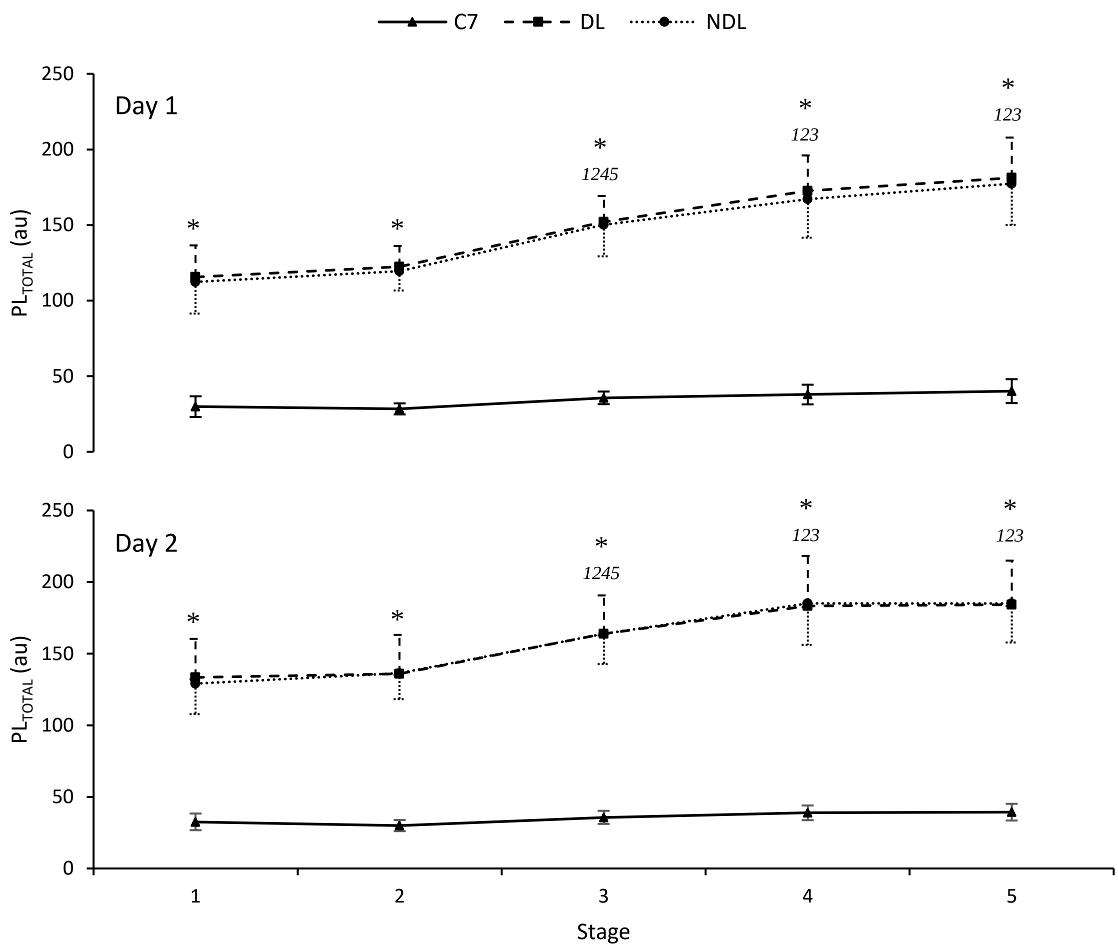

Figure 9.1. Within- and between-day PL_{TOTAL} responses to the choreographed routine at C7 and in the dominant (DL) and non-dominant (NDL). * denotes a significant main effect for unit location. [12345] signify the pairwise comparisons from stage 1 ([1]) to stage 5 ([5]). Corresponding values are mean ± σ.

A significant main effect was identified for stage ($F = 152.14$, $p < 0.01$). Post-hoc comparisons demonstrated that PL_{TOTAL} increased as a function of duration and intensity. There was no significant difference in PL_{TOTAL} between stage 5 (134.51 ± 20.55 au; CI: 126.66-142.36) and stage 4 (130.83 ± 20.23 au; CI: 124.20-137.46, $p = 1.00$), however both were significantly higher than all preceding levels ($p \leq 0.01$, $d = 0.44$-0.75). There was also a significant main effect revealed for unit position ($F = 647.75$, $p < 0.01$), with pairwise comparisons highlighting significantly lower PL_{TOTAL} at C7 (34.83 ± 5.06 au; CI: 32.68-36.99) than both the dominant (154.47 ± 23.56 au; CI: 147.60-161.34, $p < 0.01$, $d = 0.96$) and non-dominant (152.61 ± 22.44 au; CI: 142.37-162.86, $p < 0.01$, $d = 0.96$) limb, but with no significant bilateral difference between the lower limbs ($p = 0.97$). The stage *x* unit position interaction was also significant ($F = 67.01$, $p < 0.01$), with the difference between C7 and the lower limbs becoming more marked during the later stages (3-5) of the routine (see Figure 9.1).

Load Responses – Relative Uni-axial Contributions

Uni-axial load contributions to PL_{TOTAL} are quantified and displayed in Table 9.2. Results from the repeated measures ANOVA demonstrated that there was no significant main effect for day ($F = 0.43$, $p = 0.53$), stage ($F = 0.61$, $p = 0.66$), or unit position ($F = 0.32$, $p = 0.73$). However, there was a significant main effect identified for uni-axial contribution ($F = 1137.32$, $p < 0.01$), with PL_V (41.07 ± 1.92 %; CI: 40.65-41.49) representing a significantly larger contribution to PL_{TOTAL} compared with PL_{AP} (27.40 ± 1.45 %; CI: 27.12-27.67, $p < 0.01$, $d = 0.97$) and PL_{ML} (31.54 ± 1.52 %; CI: 31.10-31.97, $p < 0.01$, $d = 0.94$), whilst PL_{AP} was significantly ($p < 0.01$, $d = 0.81$) lower than PL_{ML}. There were also significant interactions highlighted for day *x* uni-axial contribution ($F = 9.50$, $p < 0.01$), stage *x* unit position ($F = 2.13$, $p = 0.44$), stage *x* uni-axial contribution ($F = 76.12$, $p < 0.01$), unit position *x* uni-axial contribution ($F = 545.95$, $p < 0.01$), and stage *x* unit position *x* uni-axial contribution ($F = 24.40$, $p < 0.01$).

Table 9.2. Uni-axial contributions to PL_{TOTAL} across stages during the ballet specific choreograph. Values are mean ± σ.

Stage	Unit location	PL_{TOTAL}		PL_{AP} %		PL_{ML} %†		PL_{V} %†	
		Accelerometer Metric							
		Day 1	Day 2	Day 1	Day 2	Day 1	Day 2	Day 1	Day 2
1	**C7**	32.56 ± 5.83	29.87 ± 6.86	25.91 ± 2.30	25.99 ± 3.98	$15.78^{\delta\eta}$ ± 1.57	$16.56^{\delta\eta}$ ± 1.87	$58.31^{*\eta}$ ± 3.33	$57.45^{*\eta}$ ± 4.49
	DL	133.42 ± 26.92	115.56 ± 20.97	27.09 ± 1.01	28.11 ± 1.54	36.17^{δ} ± 2.51	36.65^{δ} ± 1.98	36.73^{δ} ± 2.73	35.23^{δ} ± 1.87
	NDL	129.10 ± 21.33	112.37 ± 20.94	27.09 ± 1.77	27.90 ± 1.87	35.54^{δ} ± 2.00	36.71^{δ} ± 2.18	37.37^{δ} ± 2.20	35.40^{δ} ± 1.54
2	**C7**	29.94 ± 3.87	28.40 ± 3.65	25.73^{η} ± 2.31	24.99^{η} ± 1.72	$20.74^{\delta\eta 1}$ ± 1.85	$21.61^{\delta\eta 1}$ ± 2.01	$53.53^{*\eta 1}$ ± 2.65	$53.40^{*\eta 1}$ ± 2.90
	DL	136.03 ± 27.11	122.50 ± 13.60	26.66 ± 0.69	27.79 ± 0.67	36.54^{δ} ± 1.12	36.80^{δ} ± 1.97	36.80^{δ} ± 1.39	35.41^{δ} ± 1.90
	NDL	136.46 ± 18.23	119.51 ± 12.84	26.74 ± 1.07	27.20 ± 1.05	35.22^{δ} ± 1.05	36.59^{δ} ± 1.74	38.05^{δ} ± 0.69	36.81^{δ} ± 1.30
3	**C7**	35.69 ± 4.57	35.69 ± 4.18	25.14^{η} ± 1.70	25.22^{η} ± 1.02	$22.25^{\delta\eta 12}$ ± 1.48	$22.56^{\delta\eta 12}$ ± 1.42	52.61^{*1} ± 2.35	52.21^{*1} ± 2.14
	DL	163.77 ± 26.92	152.12 ± 17.08	26.93 ± 0.56	28.61 ± 0.93	36.33^{δ} ± 1.18	36.13^{δ} ± 1.43	36.74^{δ} ± 1.37	35.25^{δ} ± 1.63
	NDL	163.96 ± 21.25	150.25 ± 20.95	28.49^{2} ± 1.46	28.61^{2} ± 1.25	$34.48^{\delta 2}$ ± 1.14	$35.68^{\delta 2}$ ± 1.62	$37.03^{\delta 2}$ ± 1.08	$35.71^{\delta 2}$ ± 1.22
4	**C7**	38.87 ± 5.14	37.89 ± 6.50	24.87^{η} ± 2.23	24.68^{η} ± 0.98	$24.69^{\eta 123}$ ± 1.24	$25.14^{\eta 123}$ ± 1.45	50.44^{*123} ± 2.51	50.18^{*123} ± 2.13
	DL	183.40 ± 34.85	172.65 ± 23.38	28.54^{23} ± 1.38	29.56^{23} ± 1.08	36.80^{δ} ± 1.20	36.79^{δ} ± 1.60	34.65^{*23} ± 0.89	33.64^{*23} ± 1.72
	NDL	185.00 ± 28.85	167.08 ± 25.48	29.24^{2} ± 1.42	29.28^{2} ± 1.24	$35.60^{\delta 3}$ ± 1.12	$37.18^{\delta 3}$ ± 1.56	35.16^{*123} ± 1.25	33.54^{*123} ± 1.58
5	**C7**	39.33 ± 5.83	40.12 ± 7.91	25.11^{η} ± 1.58	25.14^{η} ± 1.02	$25.54^{\eta 1234}$ ± 1.31	$25.77^{\eta 1234}$ ± 1.32	49.34^{*1234} ± 2.01	49.08^{*1234} ± 1.80
	DL	184.24 ± 30.73	181.19 ± 26.63	30.01^{1234} ± 1.43	30.52^{1234} ± 1.19	36.76^{δ} ± 1.37	37.16^{δ} ± 1.45	33.23^{*1234} ± 1.59	32.33^{*1234} ± 2.19
	NDL	185.05 ± 27.23	177.33 ± 27.26	30.27^{1234} ± 1.17	30.43^{1234} ± 1.99	35.67^{δ} ± 0.95	36.60^{δ} ± 0.98	34.06^{*123} ± 1.47	32.96^{*123} ± 1.81

C7, 7th cervical vertebrae; DL, dominant limb; NDL, non-dominant limb, PL_{Total} (accumulated load); PL_{AP} %, percentage contribution of anterior-posterior acceleration; PL_{ML} %, medial-lateral acceleration; PL_{V} %, vertical acceleration. † denotes a significant between-day difference. $^{\eta}$ denotes a significant difference between C7 and the lower limbs locations. * denotes a significant difference compared with PL_{AP} % and PL_{ML} %. $^{\delta}$ represents a significant difference between PL_{AP}. ^{12345} signify the pairwise comparisons from stage 1 (^{1}) though to stage 5 (^{5}).

Post-hoc analyses following the significant day *x* uni-axial contribution ($F = 9.50$, $p < 0.01$) revealed that PL_{ML} (31.21 ± 1.40 %; CI: 30.83-31.59) on day one was significantly lower than on day two (31.86 ± 1.64 %; CI: 31.23-32.50, $p = 0.03$, $d = 0.21$) irrespective of stage and unit position. There was a compensatory increase in PL_{V} which was significantly higher on day one (41.60 ± 1.83 %; CI: 41.04-42.16) than on day two (40.54 ± 2.01 %; CI: 40.11-40.96, $p < 0.01$, $d = 0.27$).

The significant unit position *x* uni-axial contribution interaction ($F = 545.95$, $p < 0.01$) demonstrated that PL_{AP} at C7 (25.28 ± 1.88 %; CI: 24.75-25.80) was significantly lower than both the dominant (28.38 ± 1.05 %; CI: 28.05-28.71, $p < 0.01$, $d = 0.71$) and non-dominant (28.52 ± 1.43 %; CI: 27.89-29.16, $p < 0.01$, $d = 0.70$) limb. There was no significant bilateral difference ($p = 1.00$). PL_{ML} was also significantly lower at C7 (22.07 ± 1.55 %; CI: 21.22-22.92) compared with the dominant (36.61 ± 1.58 %; CI: 36.10-37.13, $p < 0.01$, $d = 0.98$) and non-dominant (35.93 ± 1.43 %; CI: 35.19-36.67, $p < 0.01$, $d = 0.98$), with no significant bilateral asymmetry ($p = 0.19$). PL_{V} at C7 (52.66 ± 2.63 %; CI: 51.52-53.79) was significantly higher than the dominant (35.00 ± 1.72 %; CI: 34.34-35.67, $p < 0.01$, $d = 0.97$) and non-dominant limb (35.55 ± 1.41 %; CI: 34.86-36.24, $p < 0.01$, $d = 0.97$), whilst no significant bilateral difference was observed ($p = 0.74$).

The repeated measures ANOVA identified a significant stage *x* unit position ($F = 2.13$, $p = 0.44$), stage *x* uni-axial contribution ($F = 76.12$, $p < 0.01$), and stage *x* unit position *x* uni-axial contribution interaction ($F = 24.40$, $p < 0.01$). Post-hoc analyses showed that at C7, PL_{AP} remained constant throughout the duration of the protocol ($p \geq 0.23$), whereas PL_{ML} increased significantly between stages ($p \leq 0.03$). A compensatory trend was observed or PL_{V}, with significant progressive decreases demonstrated as a function of exercise duration ($p \leq 0.01$) with the exception of stage 2 to stage 3 ($p = 0.18$). In the dominant limb, PL_{ML} did not significantly alter during completion of the ballet-specific choreographed routine ($p \geq 0.65$). PL_{AP} increased significantly from stage 4 onwards ($p \leq 0.01$), PL_{V} significantly increased from stage 3 onwards ($p \leq 0.01$), and PL_{AP} and PL_{V} in stage 5 was significantly greater compared with all other stages ($p \leq 0.01$). In the non-dominant limb, PL_{AP} increased significantly from stage 2 onwards ($p \leq 0.01$) except between stage 3 and stage 4 ($p = 0.07$). PL_{AP} in stage 5 was significantly greater than all other stages ($p \leq 0.01$). PL_{ML} was significantly greater in stage 2 ((35.91 ± 1.39 %; CI: 35.12-36.69) compared with stage 3 (35.08 ± 1.38 %; CI: 34.21, $p = 0.04$, $d = 0.29$). PL_{V} significantly increased temporally from stage 2 ($p \leq 0.04$) except between stage 4 and 5 ($p = 0.08$), and PL_V was significantly greater in stage 5 compared with all other stages except stage 4 ($p \leq 0.02$).

Physiological Responses

HR and RPE responses to the ballet-specific choreography are presented in Table 9.3. There were no significant between-day differences for HR ($F = 0.43$, $p = 0.53$) or RPE ($F = 1.07$, $p = 0.33$), nor a significant day *x* stage interaction for HR ($F = 1.98$, $p = 0.09$) and RPE ($F = 0.52$, $p = 0.79$). However, a significant main effect for stage showed that both HR ($F = 193.78$, $p < 0.01$) and RPE ($F = 113.94$, $p < 0.01$) increased significantly as function of exercise duration and intensity.

Table 9.3. Temporal effects on HR and RPE during the ballet-specific protocol. Values are mean ± σ.

	Heart Rate		Rating of Perceived Exertion	
Stage	**Day 1**	**Day 2**	**Day 1**	**Day 2**
1	161.10 ± 20.91	158.00 ± 25.63	12.90 ± 2.92	11.80 ± 3.33
2	171.90 ± 23.00[1]	170.90 ± 21.66[1]	14.60 ± 2.84[1]	13.80 ± 3.19[1]
3	181.40 ± 14.27[12]	181.00 ± 14.95[12]	16.50 ± 2.22[12]	15.80 ± 1.99[12]
4	184.50 ± 9.41[12]	185.50 ± 16.20[12]	17.30 ± 1.70[123]	17.20 ± 1.81[123]
5	178.90 ± 14.79[123]	187.90 ± 13.19[123]	18.50 ± 1.18[1234]	18.50 ± 1.08[1234]

[12345] denotes a significant difference between stages from stage 1 ([1]) through to stage 5 ([5]).

9.4 Discussion

The primary aims of the current investigation were to quantify the within- and between-day load responses to a ballet-specific testing protocol using tri-axial accelerometry, and, to examine the influence of unit location with specific focus on bilateral loading asymmetries in ballet dancers.

The first key finding demonstrated that within-day accelerometer-derived load responses to a novel ballet-specific choreographed routine increased as a product of time and intensity of exercise. PL_{TOTAL} demonstrated a progressive increment over completion of the testing protocol, a trend which is consistent with the findings of previous research conducted during the DAFT protocol (Brogden et al., 2018). Increases in accelerometer-derived load followed the same temporal pattern to those quantified for HR and RPE, thereby suggesting that resultant load responses reflected the physiological and perceptual rigours of the current testing protocol. Findings from the relative contributions to PL_{TOTAL} revealed a significant stage *x* uni-axial contribution interaction. During stage 1, the ratio of accelerations in PL_{AP}, PL_{ML} and PL_{V} was 27:30:43, which by stage 5 had altered to 29:32:39, demonstrating a similar trend to that observed during the DAFT protocol of a previous study (Brogden et al., 2018). The higher contribution from the vertical plane may be indicative of the number of aerial displacements (e.g. jeté, temps levé) included in the ballet protocol, and reflects the planar contributions to movement execution observed in study 4. That compensatory increases in PL_{AP} and PL_{ML} for a reduction in PL_{V} in stage 5 were apparent despite the greater number of jump landing manoeuvres, suggests that an altered movement strategy was employed. The temporal increases in PL_{TOTAL} may, in part, be explained by the progression of movement complexity and tempo as exercise duration increased. However, alterations in the contribution to overall movement from each plane may be symptomatic of fatigue-induced adjustments in technique and posture when performing ballet-specific manoeuvres and/or routines, which may have implications for injury risk (Barrett et al., 2016).

A second key finding from the current study highlighted a significant day *x* uni-axial contribution interaction, with relative contributions in PL_{ML} increasing on day 2 to compensate for a lower contribution in PL_{V}. This finding may be symptomatic of a change in movement economy between bouts of the piece. The experimental protocol required the same 20-minute choreographed routine to be completed twice, with 24 hours separating the two performances. There was no significant change in PL_{TOTAL}, or indeed, the physiological or perceptual responses between days, and thus, the difference in planar contributions suggests movement variability and an altered loading response between bouts. The reduced proportion of PL_{V} may be a residual fatigue response as dancers attenuate their commitment to jump landings in an effort to lower resultant impact forces. This observed modification in movement strategy may have implications for performance aesthetics and training workload prescription. Ballet dancers expend numerous hours into developing, learning and perfecting routines, and may be required to execute the complex and mechanically demanding movement profile of ballet multiple times a day, across several days of the weeks (Wyon and Koutedakis, 2013). The injury risk when performing repeated jump landing manoeuvres of high kinetic demand is magnified considering the limited time designated for recovery during a typical workday (Twitchett et al., 2010). The acute changes in mechanical load in this study oppose the findings from other research of a similar experimental design (Page et al., 2017), and also support more longitudinal observations of training and performance

(Bowen et al., 2017). The workload-injury model has demonstrated that high acute:chronic workloads are significantly associated with an increased risk of injury (Colby et al., 2014; Bowen et al., 2017). With the majority of injuries in ballet attributed to overuse (Smith et al., 2015), implementing the acute:chronic workload monitoring in ballet with emphasis on loading strategy responses to similar routines, may reveal key injury aetiology information.

A further key finding from the study was the significant main effect for unit location on PL_{TOTAL} and the relative uni-axial contributions. Accelerometers quantify resultant accelerations solely on the segment to which they are attached. Hence, it is plausible to suggest that a single unit to measure whole-body accelerations is inadequate given the multi-segment requirements of most movements (Nedergaard et al., 2017). PL_{TOTAL} showed an increase proportionate to the duration and intensity of exercise, with more profound increases observed in the lower limbs. At C7, PL_{TOTAL} was 4.6 and 4.56 times lower compared with the dominant and non-dominant limb respectively. In addition, further analysis revealed that PL_{TOTAL} at C7 increased by approximately 27% between stage 1 and stage 5, which was markedly lower than the increases observed in the dominant (~47%) and non-dominant (~50%) limbs. The symmetry in load responses demonstrated in the lower limbs may reflect the emphasis on equal limb contribution and control during development which typically begins at an early age (Bronner & Ojofeitimi, 2006).

The ratio of the uni-planar contributions to PL_{TOTAL} at C7 (25:22:53) contrasted to that of the dominant (28:37:35) and non-dominant (28:36:36) limb, which had almost identical proportions. A greater contribution of vertical accelerations at C7 may be indicative of the number of vertical displacements within the ballet-specific choreographed routine, whereas the higher relative contribution of PL_{AP} and PL_{ML} observed in the lower limbs may be explained by the technique required to execute the discrete intricacies of ballet routines. For example, the jeté requires a forward split formation of the legs which is likely to result in greater anterior-posterior accelerations at the lower limbs. Similarly, the échappé dictates a lateral displacement of the legs about a relatively stable trunk, which may produce greater medial-lateral accelerations. Another observation was the relatively negligible temporal effect on uni-axial load ratios in the lower limbs at stage 1 (28:36:36) through to stage 5 (30:37:33), compared with C7 (26:16:58 vs 25:26:49). At C7, the compensatory increases in PL_{ML} following progressive reductions in PL_{V} may be a fatigue response to increased intensity and duration.

The inter-site differences in PL_{TOTAL} and the corresponding relative planar contributions demonstrate that tri-axial accelerometry is sensitive to the individual segmental kinematics that contribute to movement in ballet. These findings corroborate those of previous studies in cricket (Greig and Nagy, 2017), dance (Brogden et al., 2018) and treadmill running (Barrett, Midgley and Lovell, 2014). Hence, quantifying accelerometer-derived load using the conventional C7 placement may underestimate the interpretation of performance rigours. The inclusion of additional units in the current study provides a greater insight into performance demands conducted in an ecologically valid environment.

Findings from the tri-and uni-axial load evaluation of ballet performance in the current investigation suggests that accelerometers are sensitive to the rigours of increasing exercise duration and demand. Further, multi-site measurements of load appear to reflect the segmental contributors to overall human movement in ballet and may benefit the strategies currently adopted in the management of ballet dancer workloads. However, the application of these findings are limited to the current experimental design. Mechanical load was quantified on two consecutive days, yet, meaningful inferences on the cumulative effects of training and/or performance on accelerometer-derived load may only be achieved via longitudinal observations spanning several weeks or months. Moreover, generalising the findings of the current study to the wider ballet population is restricted by the relatively small and homogenous sample cohort used. Despite this, inclusion of tri-axial accelerometry with consideration of anatomical specific loading may enhance current load management practices in ballet. The novel methodological approach offers scope for future research to quantify the chronic temporal biomechanical implications of executing the multi-segmental manoeuvres allied to ballet performance.

9.5 Conclusion

The current investigation represents the first to quantify within- and between-day physical and mechanical load responses to a ballet choreographed routine using accelerometry. The ballet-specific protocol of the current study was designed in accordance with the constructs of the petit and grand allegro routines of ballet, highlighting novelty in research design. The between-day difference in uni-axial contributions to PL_{TOTAL} suggests an altered loading strategy to the same performance. This finding supports monitoring ballet training and performance rigours over a sustained period, which may reveal key information towards understanding the occurrence of overuse injury, and help to determine optimal training periodisation. The current investigation demonstrated that tri-axial accelerometry was sensitive to the temporal increases in mechanical demand of this ecologically valid routine, and thus, strengthens the case for its inclusion in ballet performance monitoring. Further, between-site evaluations highlighted that accelerometry is also sensitive to the segmental kinematics that contribute to discrete ballet techniques. This finding supports the inclusion of anatomical-specific loading when quantifying the mechanical stressors of ballet performance. The bilateral symmetry in load responses evident in the lower limbs suggests that ballet training facilitates equal biomechanical development in dancers. Obtaining information relating to the demands of ballet training and performance using the methodologies of the current research, may be used towards implementing appropriate work-recovery strategies in ballet to reduce injury risk, and enhance physiological and technical development.

Chapter 10. General Discussion

10.1 Synopsis

The overarching aim of this thesis was to conduct a multi-modal biomechanical investigation of ballet performance, informed by injury epidemiology and with implications for performance profiling and injury aetiology. Injury occurrence is a multi-factorial construct, and thus, a breadth of analysis tools are required to develop a greater understanding of the demands of, and biomechanical responses to, ballet performance. The thesis was designed to evolve the laboratory-based methodologies of the initial investigations, into a consideration of opportunities for field based-analyses in the later chapters. The explicit aims of the experimental studies were to: 1) Examine bilateral 3D kinematic and external kinetic responses to a battery of ballet-specific jump landing tasks; 2) Quantify the neuromuscular responses of lower limb musculature during the selected tasks; 3) Evaluate ankle eversion/inversion strength using a functionally relevant isokinetic strength testing protocol; 4) Assess the multi-planar 'load' profiles of the ballet-specific movements using tri-axial accelerometry; 5) Explore potential associations between accelerometry and kinetic metrics; 6) Investigate the utility of accelerometry to quantify acute and cumulative load responses to a choreographed ballet routine.

The experimental chapters comprising this thesis were designed in accordance with three factors; injury epidemiology, injury aetiology, and ecological validity in sports biomechanics research. The high prevalence of injury in the amateur female ballet populations (Smith et al., 2015), provided a rationale for the participant cohort used throughout the studies of the current thesis. Epidemiology observations also established that the majority of injuries are localised to the ankle complex (Smith et al., 2015), thus further informing the design and biomechanical focus of the laboratory-based studies. A review of the risk factors associated with ankle injury demonstrated a multi-factorial aetiology (Barker, Beynnon and Renstrom, 1997; Beynnon, Murphy and Alosa, 2002; Murphy, Connolly and Beynnon, 2003, Willems et al., 2005a,b; Engebretsen et al., 2010; Witchalls et al., 2012) and thus, emphasised the necessity of a multi-modal biomechanics research design. Finally, the relative limitations in ecological validity associated with laboratory-based biomechanical analyses informed the transition from the laboratory to an authentic ballet performance setting.

In addition to the experimental setting, the movement tasks comprising the methodological approach is also fundamental in conducting a sport-specific biomechanical investigation. The typical biomechanics laboratory configuration and the constraints dictated by common methodologies such as 3D motion capture and ground reaction force plates, increases the difficulty in replicating true ballet movement. Consequently, this has culminated in the majority of existing ballet investigations using generic, clinically-focused drop and/or jump landing tasks within the methodological design (Orishimo et al., 2009; Volkerding and Ketcham, 2013; Liederbach et al., 2014; Orishimo et al., 2014; Harwood et al., 2018). These movements have limited functional relevance to ballet performance, and the reduced specificity restricts the level of understanding on the biomechanical implications of ballet movement. Ballet-specific jump landing tasks have informed some biomechanical analyses (Kulig, Fietzer and Popovic, 2011; Lee et al., 2012; Mertz and Docherty, 2012; Peng et al., 2015; Jarvis and Kulig, 2016), however these have typically been considered in isolation, thereby discounting many of the techniques comprising ballet performance. Further, the saut de chat and jeté movements common to existing investigations comprise a linear, anterior translation, which negates the lateral displacement (sissonne, sissonne PDB) and rotational demands (jeté ET) characteristic of other movements. The land and hold instruction characteristic of most studies (Kulig, Fietzer and Popovich, 2011; Mertz and Docherty, 2012; Jarvis and Kulig, 2016) is markedly different to the landing component observed in ballet, where dancers are often required to perform consecutive manoeuvres linked by a land and transition strategy. A novel and bespoke testing battery comprising seven, ballet-specific jump landing tasks of varying planar demand, was developed and subjected to a multi-modal analysis. The movement battery therefore fits the narrative of classic biomechanics tests, and the discrete techniques of ballet performance.

The specific ballet manoeuvres forming the experimental testing protocol of study 1 were chosen to differentiate between landing and hold vs landing into transition, planar/linear vs rotational flight demand, and, single- vs double-footed landings. The findings from study 1 highlighted that task variation influenced resultant kinematic and kinetic response. Evident in the joint kinematic data was an issue in coronal plane loading, with ankle joint eversion increasing over the duration of the stance phase to provide joint stability, but the degree of eversion was proportionate to the kinetic demand of the task. For instance, lower eversion was observed during the échappé in comparison with the jeté ET. Movements with a landing characterised as hold and transition such as the jeté step and sissonne PDB demonstrated more profound ankle eversion compared with their hold equivalents. This perhaps reflected an anticipatory movement towards inversion during the terminal stages of the stance phase. The apparent shift towards inversion and plantarflexion to initiate the subsequent swing phase, more closely reflects the mechanism associated with ankle injury, and thus, provides implications on the selection of tasks used in biomechanical investigations. Kinetic responses were also sensitive to the planar characteristics and technical requirements of each task, with a hierarchical ordering of movement difficulty becoming apparent. Collectively, the jeté and variants thereof (step and ET), elicited the greatest vertical ground reaction force magnitudes. These unilateral tasks comprise an approach phase, and have a greater demand for vertical displacement which likely influences the kinetic and resultant kinematic responses (Grabowski and Kram, 2008). In comparison, the échappé involved a double-leg execution, and the planar displacement in the vertical direction from a static position qualifies the lower kinematic and kinetic demand. The land and transition instruction had little bearing on the kinetic response, which contrasts with observations from the joint kinematic data.

Bilateral responses were also considered in acknowledgement of the asymmetric movement profile of ballet. Limb dominance was determined *a priori* using the preferred limb on which to execute a unilateral, ballet-specific jump landing task (Mertz and Dochery, 2012). The rationale for utilizing this method of classifying limb dominance is supported further by existing literature. (Carcia et al, 2019). Findings from the specified study observed in their application, a difference between kicking a ball and landing from a jump amongst biomechanical outcome measures. It appears therefore, that limb dominance is task-dependent and potentially cohort-specific. Given that a ballet routine comprises many jump-landing tasks, and that dancers may have a preferred limb on which to execute, the definition used in the current study is appropriate for the methodological constructs within. The results of study 1 showed no evidence of a laterality effect in either the kinematic or kinetic data, and the symmetry in biomechanical response was common to all tasks of the movement battery, irrespective of planar demand. Therefore, whilst ballet dancers may express a limb preference, there was no evidence of limb dominance. Biomechanical symmetry is crucial towards the aesthetics of ballet performance, but also in managing the risk of injury resulting from disproportionate loading to a particular limb (Zifchock et al., 2008). Ballet professionals can perhaps infer that the strategy used by female, university-level dancers to attenuate ground reaction forces is not influenced by a dominant limb. This observation may be indicative of appropriate training interventions with emphasis on limb control during ballet movement, which typically begin at an early age (Bronner and Ojofeitimi, 2006).

These strategies may facilitate chronic beneficial adaptations, with clinical implications purporting a reduced susceptibility to injury.

Having established a task characteristic hierarchy regarding the biomechanics of ballet movement in study 1, the electromyographic responses to the same battery of tasks, and using the same female dancer cohort, were investigated in study 2. EMG has been utilised previously in ballet studies, primarily to quantify the muscular contributions to discrete floor tasks (Krasnow et al., 2012; Zaferiou et al., 2017), but has yet been included to supplement the kinematic and kinetic responses to the same movements. The functional kinesiology of the ankle determined from the kinematic data of study 1, dictated the selection of the muscles used in the EMG analysis. The amplitudes in EMG response mirrored the hierarchical ordering of task mechanics within the movement battery, and individual muscle activity reflected functional kinesiology and the magnitudes of joint displacement specific to each task. For example, the échappé generated the lowest kinetic response, and observed the lowest kinematic demand regarding sagittal and coronal plane motion, and thus, had a lower EMG response. Whereas the jeté-inclusive tasks had greater biomechanical demand, reflected in a greater muscular response. The land and transition component specific to the jeté step and sissonne PDB elicited an alteration in EMG response. A significant reduction in peroneus longus activity was compensated by an increase in lateral gastrocnemius activity. This may be symptomatic of a change in the responsibility of the neuromuscular system, from providing full support during body weight stabilisation, to facilitating joint co-ordination towards a subsequent ballet movement. The bespoke EMG responses to the ballet-specific movements, represents the kinetic demand and kinematic characteristics specific to each task, and the contrasting amplitudes support the classic force-EMG relationship (Roberts and Gabaldon, 2008). The lack of asymmetry in neuromuscular response is consistent with the kinematic and kinetic observations of study 1, and indicates that a similar motor control strategy is used to govern ankle joint function during ballet movement. This observation may again reflect a training adaptation, thereby supporting existing interventions in ballet towards bilateral technical development.

Joint strength not only plays an important role in force generation capacity towards ballet movement and performance, but also in joint kinematics to attenuate the high ground reaction forces associated with specific jump landing tasks. Study 3 comprised an evaluation of ankle joint strength using the same participant cohort as the preceding studies. Minimal studies have quantified ankle joint strength capacity in female dancers (Thomas and Parcell, 2004; Kenne and Unnithan, 2008), and existing methodologies have yet to consider the eversion/inversion mechanism common to both injury occurrence, and, ballet-specific movement as demonstrated in study 1. In contrast to previous isokinetic protocols, whereby angular velocities appear arbitrarily selected to represent a slow ($30°·s^{-1}$) and fast ($120°·s^{-1}$) motion (Willems et al., 2002; Pontaga, 2004), the configuration of the dynamometer and development of the isokinetic testing protocol in study 3 was functionally relevant, supported by the kinematic data of study 1. Specifically, sagittal and coronal plane joint displacement was used to orientate the foot attachment in partial plantarflexion, and, to mediate the experimental movement range in eversion and inversion. Further, joint angular velocity data in the coronal plane demonstrated that the ankle moves at $\leq 120°·s^{-1}$ for the majority and up to 89% of the stance phase of all ballet tasks, thus supporting the inclusion of a range of angular velocities in the isokinetic testing protocol of study 3.

The findings of study 3 revealed that university-level female dancers are eccentric inversion strength dominant, with more profound strength capacity evident at angular velocities $\geq 60°·s^{-1}$, and for all angular displacements. This may have important clinical implications in preventing the inversion mechanism common to ankle injury incidence, particularly when transitioning into a connecting technique given the translation towards plantarflexion and inversion demonstrated during the jeté step and sissonne PDB. For all contraction modes, bilateral strength symmetry was evident in peak torque and the corresponding dynamic control ratios, but also in the angle of peak torque and functional range. Analysing strength capacity beyond the maximum value is important to the understanding of how strength is maintained over an isokinetic range. The demonstrable equivalence in bilateral limb strength may reflect early prescription and technical adaptations, with a training emphasis on equal limb input when executing the movements comprising ballet performance. EMG responses to the strength assessment were also quantified in consideration of the neuromuscular system’s role in force production. EMG data revealed a bilateral symmetry response in support of the observations in strength, and, the EMG responses to ballet-specific movement as quantified in study 2. Rather than normalising to an MVIC, the neuromuscular strategy towards maximal strength testing was discussed in relation to the functional requirements of ballet movement. A comparison of the respective EMG responses demonstrated that maximal isokinetic strength testing elicited significantly lower magnitudes than when performing the movement battery. This information underlines the biomechanical demands of ballet and associated risk of injury when performing the explosive jump landing tasks, reflected in injury epidemiology. Moreover, the data strengthens the argument for alternative methods of normalising EMG, and questions the validity of an MVIC technique for functional movement tasks.

Whilst studies 1-3 were designed with greater specificity towards ballet movement, and adopted a multi-modal biomechanical approach, they were conducted in a laboratory setting. Data collection tools in 3D motion capture, ground reaction force plates, and isokinetic dynamometers have limited portability beyond this environment. The call for alternative methods towards biomechanical assessments in more ecologically valid settings was acknowledged, and study 4 therefore explored the utility of accelerometry to quantify mechanical load responses to the same movement battery. Accelerometry enables a 3D evaluation akin to force plate and motion capture analyses, and the relatively high sampling frequency suits the explosive jump landing manoeuvres typical of ballet.

The laboratory-based studies discussed above were designed and conducted in accordance with the biomechanical demands of ballet, with an appreciation of improved ecological validity throughout. Whilst study 4 highlighted the potential for accelerometry to quantify movement with comparable sensitivity to the other data collection modes used, investigating a potential method to quantify external load in applied settings was an important endeavour in the progression towards field-based assessments of ballet performance. Study 4 revealed that accelerometer responses reflected the movement characteristics and biomechanical response specific to each jump-landing task. Specifically, the greater magnitudes in PL_{TOTAL} during the jeté-inclusive tasks corresponded with kinetic and kinematic demands of the same movements in study 1. The same hierarchical ordering of movement demand demonstrated in study 1 was further supported by the lower PL_{TOTAL} values overserved in the échappé, sissonne, and sissonne PDB techniques respectively. Planar contributions to PL_{TOTAL} were also quantified to provide a greater level of understanding into how accumulated load is derived. The finding of a greater input of PL_{V} was anticipated, and perhaps reflects the aerial displacement typical of each task. A greater contribution in PL_{ML} over PL_{AP} was observed for all movements except the jeté ET, which may reflect the direction of movement when performing ballet manoeuvres. However, it may also provide an indication on the strategy used to provide whole-body stability during landing. This notion was particularly evident during the échappé, which observed a greater contribution from PL_{AP} compared with the other movements. The échappé is characterised by a position of turnout, which induces pronation of the foot and creates resultant instability in the anteroposterior plane (Carter, Bryant and Hopper, 2019). The greater contribution in PL_{AP} therefore, appears to reflect a compensatory mechanism to provide stability on landing.

Planar accelerations were examined to provide insight on the direction of movement in each plane. In the anteroposterior plane, backwards accelerations were three times greater than forwards acceleration, potentially reflecting trunk displacement in providing stability when arresting forward momentum. There was no significant difference in mediolateral movement, which further strengthens the suggestion that female dancers adopt an anteroposterior stabilisation strategy when landing from a ballet-specific jump. The findings from study 4 highlighted the potential use of accelerometry for valid biomechanical assessments of ballet movement, with comparable sensitivity to the data collection modes used in the preceding chapters. Further research on the efficacy of accelerometry to quantify external load in applied settings is advocated in the progression towards biomechanical assessments in ballet training and competition.

The aim of study 5 was to explore potential associations between accelerometry and kinetic parameters. The relationship existing between accelerometry and vertical ground reaction force has been investigated in general locomotive tasks (Wundersitz et al., 2013; Simons and Bradshaw, 2016; Nedergaard et al., 2017; Edwards et al., 2019), with the efficacy of accelerometry to estimate GRFs suggested to be task specific. This relationship in ballet movement has been quantified in a single previous study using a solitary technique only (Almonroeder et al., 2019). The contrasting evidence within relevant studies, and the limited information available from a ballet context, highlighted the need for further investigation across a range of ballet tasks. A secondary analysis was conducted on the kinetic data obtained in study 1 and the accelerometer-derived loads quantified in study 4.

The findings from study 5 revealed strong, positive correlations between impact peak vertical accelerations and vertical ground reaction forces for the échappé ($r = 0.59$), sissonne PDB ($r = 0.59$), jeté step ($r = 0.65$), jeté ET ($r = 0.63$) and sissonne ($r = 0.69$), and very strong, positive correlations for the temps levé ($r = 0.77$) and jeté ($r = 0.81$). Further, the relationship coefficient between impact peak vertical accelerations and mean vertical loading rates was strong, positive for the sissonne PDB ($r = 0.59$), and very strong, positive for the temps levé ($r = 0.75$), jeté ($r = 0.82$) and jeté ET ($r = 0.84$). The contrasting coefficients appear to reflect the degree of vertical displacement characteristic of each task. For example, the jeté and temps levé movements have a greater requirement for vertical displacement compared with movements with a lateral translation (sissonne), a rotational flight (jeté ET), or a transition landing component (jeté step, sissonne PDB). That only vertical accelerations were considered in the accelerometer-ground reaction force relationship may explain the variance in correlation coefficients between tasks. The strength of this correlation may also be attributed to the conventional accelerometer placement at C7. This location is recommended by the manufacturer to enhance satellite signalling for the GPS component; however, accelerometers are sensitive to the segment on which they are attached. Hence, positioning the device at C7 may limit the interpretation of movement and resultant loads experienced during performance. Nevertheless, the findings from study 5 suggest that accelerometry can provide a valid indication of the external forces imposed by ballet movement, and advocate the use of accelerometry to complement existing methods of biomechanical assessment in the field. In-vivo measures of ballet performance is achievable with accelerometry, providing an avenue to quantify training and competition loads from a biomechanical perspective, and enabling ballet to progress towards the level of athlete monitoring observed in other team invasion modes.

Study 6 represented the final experimental chapter of the current thesis, and utilised accelerometry to assess load responses to a choreographed ballet routine, whilst also considering anatomical location in the interpretation of movement demand. In addition, biomechanical responses were quantified during two experimental trials separated by 24 hours, as to partially simulate the congested scheduling typical of ballet training and competition. The choreographed performance piece was designed in accordance with the multiple stage, progressive intensity principles of the dance aerobic fitness test (DAFT) (Wyon et al., 2003), but tailored to include some of the tasks comprising the movement battery of the earlier studies. The findings from study 6 demonstrated that accelerometry-derived load responses increased proportionately with routine duration and exercise intensity. Comparisons between accelerometer devices locations revealed significantly higher loading in the lower limbs compared with the C7 location, with more profound differences highlighted in the latter stages of the routine. This data suggests an interaction between fatigue and anatomical location, and the interpretation of a fatigue response may only be appropriately informed by additional units to the C7 device. The rationale for units positioned at the lower limb was developed in accordance with injury epidemiology, and therefore, unit location ought to be an important consideration in research design when quantifying load responses using accelerometry. Further evidence of a potential fatigue effect was evident in the significant day *x* uni-axial contribution. Analyses highlighted a compensatory increase in PL_{ML} on day 2 to account for reductions in PL_V, which may be indicative of a lower commitment to the aerial component of ballet jump-landing tasks, with implications for ballet performance and injury risk. The findings from study 6 support the use of accelerometry in monitoring the biomechanical demands of ballet training and performance. Subsequent information may be used to inform work-recovery strategies to reduce injury risk and enhance performance, but requires a consideration of accelerometer location in the interpretation of load response.

10.2 Recommendations for Future Research

Caution should be taken when interpreting and generalising the findings beyond the specific cohort used. In many instances, comparable literature is not available to critically discuss or indeed contextualise the findings of the current thesis. Further research is required to substantiate the observations contained within, and, the following section provides recommendations on how this research domain may be developed in future biomechanics investigations.

Participants

Although justified by injury epidemiology data, the amateur, adult female ballet dancer cohort used restricts the interpretation of the findings to this specific population. Hence, the extent to which they apply to young/adolescent dancers, male dancers, professional populations, and indeed those training in other dance genres, remains largely unexplored. The current thesis observed a level of bilateral symmetry in amateur dancers, and intuitively, this would be apparent in professional cohorts too. However, what remains unknown is the stage of maturation at which symmetry is developed. Ballet training typically begins from young age, with technique classes assumed to prescribe tasks with equal input from both limbs given the kinaesthetic demands of ballet performance. In other asymmetric kinetic sports such as soccer, young athletes often have a dominant limb to execute dynamic movements requiring strength (Fousekis, Tsepis and Vagenas, 2012). This has been shown to induce a leg asymmetry in change of direction tasks inherent to these sports (Rouissi et al., 2016). Future research investigating the level of bilateral (a)symmetry in ballet dancers at different stages of biological development, may highlight important training implications, particularly towards prescription in sports were asymmetry is apparent. The classification of limb dominance in the current thesis was defined prospectively, using the preferred limb on which to land from a unilateral, ballet-specific jump landing task (Mertz and Docherty, 2012). With regards to sports performance, it is important to consider and distinguish laterality (skill dominance) from force dominance (the limb with superior force capacity) (Maloney, 2019). For example, Carcia, Cacolice and McGeary, (2019), highlighted in their application of limb dominance classification, a difference between kicking a ball and landing from a jump. Defining limb dominance therefore, appears to be task-dependent, and these observations should be considered in the limb dominance classification methods of future research studies.

Strict inclusion and exclusion criteria dictated that participants were injury free. However, epidemiology data suggests that ballet dancers sustain injury at a high frequency, which may induce a limb dominance effect and resultant limb asymmetry. The current thesis demonstrated that bilateral symmetry is evident in non-injured dancers at the time of training and competition, reflected in a range of biomechanical factors such as kinetic and joint kinematics, neuromuscular response and isokinetic strength. Hence, future research investigating differences in symmetry response in a previously injured dance group, or indeed a case study monitoring the rehabilitation of a dancer post-surgery, may provide key aetiology information informing return to training, and also, towards identifying markers for initial, and recurrent, injury.

Methodological Paradigm

In study 1, a rigid segmental foot model was utilised to quantify ankle joint kinematic responses to ballet specific movement. As shown in the injury epidemiology section (chapter 2), the primary anatomical location for injury is the ankle complex. In some studies, however, ankle injuries encompassed trauma to the foot, and therefore, modelling the foot as a single, rigid segment is not suitable to quantify foot kinematics during ballet movement. Hence, a multi-segmental foot model may be required in future research to advance the understanding on the rigours of ballet movement and the ensuing risk of injury to dancers (Carter, Bryant and Hopper, 2019). However, these observations also underline the importance of quantifying specific injury diagnoses in epidemiology research, thereby ensuring that the methodological approaches of subsequent studies comprise the most appropriate means of investigating injury risk.

Another noteworthy consideration from study 1 towards future research concerns jump height and approach velocity. These variables influence the magnitude of GRFs (Grabowski et al., 2008), and intuitively, the resultant joint kinematics to attenuate the external loads imposed by movement. It is therefore recommended that future studies account for these variables to strengthen the understanding on the mechanical demands of ballet, and the technique employed by dancers in response. The simultaneous use of 3D motion capture and force plate analysis enabled a range of kinematic and kinetic responses to ballet-specific movement to be quantified. However, these methodologies also facilitate inverse dynamics analysis, providing information on joint-specific loading. The calculation of joint moments would allow the force acting about the foot and ankle complex to be explored, potentially advancing the understanding of the mechanical rigours of ballet movement and the associated risk of injury.

Neuromuscular responses to ballet movement were quantified in study 2, with mean and peak amplitudes quantified. That the movement battery was completed in a non-fatigue state, prohibited a meaningful consideration of frequency domain parameters. With a high proportion of injuries attributed to overuse, monitoring EMG responses over a prolonged ballet exposure would enable the effects of fatigue to be explored. Future research may indeed benefit from the portability offered by contemporary EMG technologies. The implications of ballet performance fatigue on potential alterations in the frequency of neuromuscular activity, may highlight key information towards understanding the risk of injury.

The isokinetic dynamometer configuration used in study 3 was supported by the joint kinematic data collected in study 1, resulting in the foot attachment being placed in 20° of plantarflexion to partially replicate landing mechanics. The effect of this modified set up on resultant isokinetic strength remains unclear without consideration of the conventional neutral position, or indeed other joint displacements. The joint kinematic data in study 1 demonstrated that the ankle moves through an angular displacement in the sagittal plane, with displacement demonstrating a task-dependent response. The isokinetic strength analysis of study 3 comprised a consideration of angle-specific torque in eversion/inversion, and a similar approach may be adopted in the dynamometer configuration of future studies. Indeed, it may be important to evaluate ankle eversion/inversion strength capacity using multiple foot attachment displacements. Whilst future studies will be prohibited from examining eversion/inversion strength in full plantarflexion – typical of the articular adaptations in ballet dancers – due to ethical constraints, perhaps isokinetic strength can be evaluated up to, and beyond, the 20° angle used in the current thesis.

Associations between Lab- and Field-based Measures

Association between the data collected in accelerometry and GRF analysis was explored in study 5. Although strong, positive correlations were identified between accelerometry and GRFs, and, the corresponding $_V$MLR across the movement battery, the strength of the relationship appeared to reflect the degree of vertical displacement specific to each task. This generally reflects observations from the wider literature base during general locomotor tasks, in which the vertical acceleration-vertical ground reaction force relationship has demonstrated contrasting evidence during running (Wundersitz et al 2013; Nedergaard et al., 2017; Edwards et al., 2019), but is strong during jump-landing tasks (Simons and Bradshaw, 2016; Setuain et al., 2016). In addition, that the coefficient value (r) was less than one for all tasks, and lower in the movements comprising a lateral or rotational element, suggests that other factors influence the accelerometry-GRF beyond the vertical component. Future research may investigate this relationship using the tri-axial capacity of accelerometry. Consideration of anteroposterior and mediolateral acceleration may improve the accelerometry-GRF correlation particularly in movements with a multi-planar kinematic profile. Subsequent research may advance the utility of accelerometry for estimating the external force imposed by ballet movement in training and performance settings.

Movement Tasks and Choreographed Routine

A novel movement battery consisting of 7 ballet-specific jump-landing manoeuvres was designed and incorporated into the initial experimental studies. The discrete movements were selected to distinguish between landing strategy (hold vs transition), flight demand (linear vs rotational), and single- vs double-legged landing execution. Whilst the discrete techniques were sensitive enough to elicit a task-dependent biomechanical response, the ballet performance repertoire contains many more movements than those used in the current thesis. Hence, a true understanding on the complexity and mechanical demands of ballet may only be achievable with a consideration of all movements, and future studies may develop the movement battery to include more tasks.

A choreographed ballet routine was developed and used in study 6. The performance piece was constructed within the narrative of the Dance Aerobic Fitness Test (DAFT) protocol (Wyon et al., 2003), but with demonstrable ballet specificity via inclusion of some of the techniques comprising the movement battery. The DAFT protocol has been validated as a maximal aerobic test, but the movements within have limited functional relevance towards the ballet movement profile. The performance routine used in study 6 has yet to be validated, highlighting an area for future research. That said, the physiological responses to each stage of the routine were similar to those observed during the DAFT (Wyon et al., 2003; Brogden et al., 2018). Subsequent research therefore, may indeed validate the performance routine from the current thesis as a maximal performance test, to inform future biomechanics research experiments in ballet.

Study 6 provided initial insight into the mechanical responses of dancers to acute and chronic bouts of ballet performance. Monitoring training and competition workloads is a crucial consideration in periodisation, enabling dancers to perform at peak capacity whilst managing the risk of injury. Currently, data on the tolerance of dancers towards ballet training and performance is limited, and significant further research is needed to address the high prevalence of overuse injuries observed in the ballet populations (Smith et al., 2015). In other sports, governing bodies have imposed restrictions on athlete exposure to the mechanical demands of the specific movement profile. For example, fast bowlers in cricket are limited to a set number of overs for age groups up to 21 years old. These sanctions were implemented based on injury epidemiology, with data highlighting a disproportionate incidence of lumbar spine injuries in adolescent bowlers. In addition, fixture scheduling is carefully considered to prevent athletes in sports such as soccer and rugby, from competing maximally on consecutive days, thereby minimising the risk of injury attributed to overuse. In ballet however, there are no government body approved restrictions on competition frequency, despite dancers performing for many hours a day, across several days a week (Twitchett, Koutedakis and Wyon, 2009). Hence, further research is needed to inform a ballet exposure threshold that may help to reduce overuse pathologies. In study 6, accelerometer-derived load responses to a choreographed routine were quantified over two consecutive days. However, this limits the interpretation of fatigue-induced responses to ballet, and therefore, more longitudinal observations of ballet training and competitions, may be necessary to further understand the occurrence of injury. This is also supported by the workload-injury risk paradigm demonstrating an association over a 3-week period and beyond (Colby et al., 2014; Bowen et al,. 2017; Bowen et al., 2019). It is important to note that PhD research is bound by ethical considerations and time constraints, and therefore, a prolonged evaluation of ballet training and performance was not viable. Future studies may indeed apply accelerometry to examine load responses over a ballet training and competition season. This approach would also be strengthened by the recording of injuries sustained over the same period, and a retrospective analysis of accelerometry data may highlight workload and/or recovery issues leading up to injury occurrence.

10.3 Practical Implications

The findings of this PhD project have implications for the methods employed in biomechanical analyses, and, the strategies used in motoring ballet performance in applied settings. The use of a multi-modal biomechanical approach reflects the multi-factorial aetiology of injury, and the performance demands of ballet, and synergy and cohesion in the use of various data collection techniques provided the most appropriate means of quantifying the biomechanical demands of ballet, helping to achieve the initial aims of the thesis. The experimental studies collectively, provide a comprehensive profile on the biomechanical demands of ballet movement, and, the mechanical capacity of dancers in response, using an array of biomechanical analysis tools. The ballet-specific jump landing movement battery adds a novel dimension and improves ecological validity, but represents the specific movements completed during performance. The movement battery was sensitive enough to elicit a task-dependent response, providing new insight regarding the kinetic and ankle joint kinematic demands allied to a range of ballet movements. This response was reflected in a multi-modal analysis, with consistency in the main effects for task demand and limb symmetry, and the development of a hierarchical ordering of ballet movement difficulty.

The association in biomechanical response across a range of data collection tools evolved from a general interpretation, to investigating a specific correlation in accelerometry and vertical ground reaction force. The accelerometry research paradigm enables mechanical assessment in more ecologically valid environments, and in-vivo assessments of biomechanical response can be achieved in ballet training and competition. Anatomical location of the accelerometer is a key consideration in research design. Placement of the device ought to be informed by injury epidemiology, and interpretation of movement response may provide important clinical information towards understanding injury occurrence. Hence, the findings from the current thesis have an applied value, and the methodological approach comprising each experimental study offers novelty and innovation in the assessment of biomechanics in ballet. Further research is required to substantiate the findings of the current thesis, and, to develop understanding on the mechanical demands of ballet, and the implications for injury risk. Field-based performance profiling may be favourable over laboratory-based designs, particularly in addressing the high prevalence of overuse pathology observed in the ballet populations. Subsequent information may establish a mechanical tolerance to ballet exposure, which can be carefully monitored to mitigate against potential decrements in performance, and an increased risk of injury.

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Appendices

Appendix 1. The Cumberland Ankle Instability Tool (CAIT) questionnaire (Hiller et al., 2006)

Please tick the ONE statement in EACH question that BEST describes your ankles.

		LEFT	RIGHT	SCORE
1.	**I have pain in my ankle**			
	Never	☐	☐	5
	During Sport	☐	☐	4
	Running on uneven surfaces	☐	☐	3
	Running on level surfaces	☐	☐	2
	Walking on uneven surface	☐	☐	1
	Walking on level surfaces	☐	☐	0
2.	**My Ankle feels UNSTABLE**			
	Never	☐	☐	4
	Sometimes during sport (not every time)	☐	☐	3
	Frequently during sport (every time)	☐	☐	2
	Sometimes during daily activity	☐	☐	1
	Frequently during daily activity	☐	☐	0
3.	**When I make SHARP turns, my ankle feels UNSTABLE**			
	Never	☐	☐	3
	Sometimes when running	☐	☐	2
	Often when running	☐	☐	1
	When Walking	☐	☐	0
4.	**When going down the stairs, my ankle feels UNSTABLE**			
	Never	☐	☐	3
	If I go fast	☐	☐	2
	Occasionally	☐	☐	1
	Always	☐	☐	0
5.	**My ankle feels UNSTABLE when standing on one leg**			
	Never	☐	☐	2
	On the ball of my foot	☐	☐	1
	With my foot flat	☐	☐	0
6.	**My ankle feels UNSTABLE when**			
	Never	☐	☐	3
	I hop from side to side	☐	☐	2
	I hop on the spot	☐	☐	1
	When I jump	☐	☐	0
7.	**My ankle feels UNSTABLE when**			
	Never	☐	☐	4
	I run on uneven surfaces	☐	☐	3
	I jog on uneven surfaces	☐	☐	2
	I walk on uneven surfaces	☐	☐	1
	I walk on a flat surface	☐	☐	0
8.	**TYPICALLY, when I start to roll over (or 'twist') on my ankle, I can stop it**			
	Immediately	☐	☐	3
	Often	☐	☐	2
	Sometimes	☐	☐	1
	Never	☐	☐	0
	I have never rolled on my ankle	☐	☐	3
9.	**After a TYPICAL incident of my ankle rolling over, my ankle returns to 'normal'**			
	Almost immediately	☐	☐	3
	Less than one day	☐	☐	2
	1-2 days	☐	☐	1
	More than 2 days	☐	☐	0
	I have never rolled over on my ankle	☐	☐	3
	Total			

Appendix 2. Pre-exercise medical questionnaire

Department of Sport and Physical Activity

Medical Questionnaire

The information in this document will be treated as strictly confidential

UK data protection law changed on 25 May 2018 when the General Data Protection Regulation (GDPR) came into effect. The University will not disclose your personal information to a third party, without explicit consent, unless there is a legitimate reason for doing so. Disclosing information to a third party for a legitimate interest is permitted only if disclosure would not prejudice the rights and freedoms or legitimate interests of the data subject. This new legislation gives you more control over how your data is used. We have updated the University Privacy Notice to reflect these changes.

This questionnaire is for general health screening of students/laboratory participants. Members of staff responsible for supervising students/laboratory participants engaging in physical exercise during timetabled sessions and other teaching, research or work-related activity are expected to conduct further health screening as appropriate.

Name: ……………………………………………Date of Birth: ………………………

Age…………Sex:……………………………………………………………………

Height: ………………………. Body Mass: ………………………

Degree programme (if applicable)…………………………………………………

Please answer the following questions by putting a circle round the appropriate response or filling in the blank

1. How would you describe your present level of **exercise** activity?

Sedentary / Moderately active / Active / Highly Active

2. Please outline your present weekly **exercise** activity:

……………………………………………………………………………………………
……………………………………………………………………………………………
……………………

3. How would you describe your present levels of **lifestyle** activity?

Sedentary / Moderately active / Active / Highly Active

4. How would you describe your present level of fitness?

Unfit / Moderately fit / Trained / Highly Trained

5. What is your current (main) occupation/job (if appropriate)? ....

6. Smoking habits: Are you currently a smoker? Yes / No

If so, how many do you smoke? …….. per day

Are you a previous smoker? Yes / No

How long is it since you stopped? …….months/years

How many did you smoke? ..….. per day

7. Do you currently drink alcohol? Yes / No

If you answered **YES**, do you drink more than 14 units a week?

(Small glass of wine (175 ml) = 2.1 units; pint of lager (5%) = 3.0 units; single spirit = 1.4 units) Yes / No.

8. Have you had to consult your doctor for any medical reason in the last 6 months? Yes / No

If you answered **YES**, have you been advised NOT to exercise? Yes / No

9. Are you presently taking any form of medication? Yes / No

If you answered **YES**, have you been advised NOT to exercise? Yes / No

10. To the best of your knowledge do you, or have you ever, suffered from?

Diabetes Yes / No

Epilepsy Yes / No

Anaemia Yes / No

*High blood pressure Yes / No

Heart complaint Yes / No

Asthma or other lung disease Yes / No

Aneurysm / embolism? Yes / No

Marfan syndrome Yes / No

Other cardiovascular disease Yes / No

**Circle 'Yes' if you are currently taking blood pressure medication*

11. * Are you over 45, and with a history of heart disease in your family? Yes / No

12. Do you currently have any form of muscle or joint injury? Yes / No

If you answered YES, please give details: ………………………………………………………………………………………………… ………………………………………………………………………………………………… …………………………

13. Have you had to alter your normal physical activity level in the last 2 weeks? Yes / No

If you answered YES, please give details: ………………………………………………………………………………………………… ………………………………………………………………………………………………… …………………………

14. Please answer the following questions:

   a. Are you suffering from any known infection? Yes / No

   b. Have you had jaundice within the previous 12 months? Yes / No

   c. Have you ever had any form of hepatitis? Yes / No

   d. Are you haemophiliac? Yes / No

15. As far as you are aware, is there anything that might prevent from safely engaging in physical exercise as part of your studies? Yes / No

IMPORTANT: If you experience any change in your health you must inform a member of staff as soon as possible

Student signature: …………………………………………………. Date: ……………..

Staff signature: ……………………………………………………. Date: ……………..

Parent (if minor): ………………………………………………………. Date: ……………..

THE FOLOWING SECTION IS ONLY TO BE COMPLETED FOR FURTHER HEALTH SCREENING AT SUBSEQUENT TIMES, SUCH AS WHEN STUDENTS ENTER SUBSEQUENT YEARS OF STUDY OR WHEN PARTICIPANTS UNDERTAKEN REPEAT SESSIONS

Is the information you provided above still correct, and can you confirm that you have NOT experienced any new injury, illness, or other health condition?

Repeat 1

Yes / No *

Signature: Date:

Additional information required:

Repeat 2

Yes / No *

Signature: Date:

Additional information required:

Repeat 3

Yes / No *

Signature: Date:

Additional information required:

Repeat 4

Yes / No *

Signature: Date:

Additional information required:

Repeat 5

Yes / No *

Signature: Date:

Additional information required:

Appendix 3. Participant Information and Informed Consent Sheets

Project Title:

1) **The Kinematic and Kinetic Responses to Ballet-specific Movements**
2) **Electromyographic Responses to a Ballet-specific Movement Battery**
3) **Isokinetic Strength Profiling of Amateur Female Ballet Dancers**

Lay Title:

A Biomechanical Analysis of Ballet-specific Movement

You are invited to take part in a research study within the Sport and Physical Activity Department at Edge Hill University. Before you decide whether to partake, it is crucial you understand why the research is being conducted, and, what will be required from you. Please allow yourself time to read the following document carefully and discuss it with friends, family and your GP if you wish. Do not hesitate to contact me if any of the elements in the research require clarity, or simply, that you would like more information.

Purpose of the Study

Previous research has identified high injury rates in dancers. Whilst there is an understanding of the types of injuries sustained in dance, there is little evidence regarding the mechanisms (risk factors) for injury. Therefore, using a variety a testing procedures, the current research aims to investigate the mechanisms associated with an increased risk of injury to dance performers.

Eligibility

You are invited to participate in the research given your status as a currently active female dancer, and your familiarity and experience with the movements associated with dance performance. You will be between the ages of 18-25 and injury free in the six months leading up to your testing session. Participation is entirely voluntary and you will be asked to provide written informed consent should you wish to partake in the research. Participation reserves your right to withdraw from the research at any time during the experimental trials, and, up to four weeks following your final data collection session without reason, or consequence.

To determine your eligibility for the exercises in the study, you will undergo a thorough health assessments. The process will involve you completing an extended questionnaire to provide information regarding any current or previous medical issues, and a history of injuries. Should you present any medical or injury cases, you will be unable to take part in the testing procedures until further measures are applied, and in some instances, advised to seek the help of a medical practitioner. If you display, or inform the researcher of, any signs of physical or mental distress during the data collection process, the session will be terminated.

Study Procedures

If you consent to take part, you will be asked to attend the biomechanics laboratory at Edge Hill University (Wilson Centre, Department for Sport and Physical Activity), on a maximum of two occasions. Depending on the time available to you, the data collection process may be completed during a single visit. The visits will be arranged at a time and date to suit you, and the testing session(s) will consist of familiarisation, and two experimental trials. Your entire data collection process will last no longer than 2 hours.

The familiarisation process will ensure you are clear and comfortable with the equipment and exercises required during the testing session. Details of the experimental trials to which you will be familiarised with, are contained in the following sections. The experimental trails will comprise the completion of dance-specific exercises in which you will perform several jump-landing manoeuvres, and an assessment of ankle strength using an isokinetic dynamometer. A number of measures, as described in the subsequent section will be recorded during the tasks. Prior to each of the experimental trails, you will be required to complete a 10 minute warm-up comprised of general mobilisation exercise and dynamic stretching.

Global Positioning System (GPS)

During the dance-specific exercises you will be fitted with a neoprene vest containing a small matchbox-sized device, which will sit between your shoulder blades. A second device will be attached to your lower back to approximate centre of gravity and secured using under-wrap tape. The devices will record continuously throughout each trial and will be used to measure 3D accelerations as a marker of exercise 'load'. The equipment should not provide you with any discomfort or compromise movement, and therefore, the risk to you is minimal.

Electromyography (EMG)

EMG will be used to assess your muscle activity during the dance-specific exercises. Preparation will comprise shaving, cleaning and abrasion of the appropriate area of skin. Three electrodes will be placed on the Peroneus Longus, Lateral Gastrocnemius and Tibialis Anterior muscles of each leg. The electrodes will be connected to a receiver, enabling the EMG data to be transmitted for analysis. You should not feel any discomfort and the risks are suggested to be minimal.

Ground Reaction Force Analysis

You will be required to complete three trials of seven dance-specific movements; 1) Jete; 2) Jete into step; 3) Echappe; 4) Sissone; 5) Sissonne into pas de bourres 6) Temps leve: 7) Jete en tourney, for both the left and right leg. Using the appropriate technique for the identified jumps, you will initiate the jumps at a pre-determined distance from the force plate, and make contact with the force platform upon landing. Reflective tracking markers will be placed on the body for the purposes of data collection. The markers will be positioned on the ankle and on parts of the foot and knee, and, placement will be achieved by palpating (feeling) for the correct marker placement. A multi-camera motion capture system will be used to track the position of the markers during the trials, to enable the measurement of ankle joint motion. Use of the motion capture system will not make you identifiable, as the data only displays the markers located on the ankle and the foot. Hence, the motion capture system upholds your anonymity. The movements will mirror those observed during dance, and will be performed on a similar surface.

Isokinetic Dynamometry (IKD)

An isokinetic dynamometer (IKD) will be used to measure your ankle joint muscular strength. The protocol will require you to sit in the chair of the dynamometer in a comfortable position, whilst straps will keep you safely secure. Once positioned, you will be required to complete maximal muscle contractions for ankle inversion and eversion in accordance with the instructions of the researcher. For each assessment, you will be required to complete three sets of 5 repetitions at various speeds (30, 60, 90, 120 degrees/second) for each leg. You should not feel any pain during completion of the trials; however, you may feel slightly fatigue towards the end. You may have very little experience in using an isokinetic dynamometer, and, as such, you will be given sufficient time to familiarise yourself with the procedure and exercises concerned.

Other Requirements

- You must be currently active in dance, free from injury in the six months leading up to your testing session, and older than 18 years of age.
- For the purposes of data collection, you will be required to wear shorts and your footwear should match that worn during training/competition. Please be advised that appropriate facilities will be provided for any changing requirements.
- You will be required to refrain from alcohol and vigorous exercise 24 hours prior to each experimental trial.

Risk to Participants

The procedures outlined in the two experimental trials above are commonplace in sport and physical activity research. The testing exercises will be within the limits of those you experience during typical dance activity and therefore, the risk of injury is suggested to be minimal. All potential injury risks will be mitigated through medical and pre-exercise screening assessments, and you will be monitored closely for signs of pain and discomfort during completion of the exercise. You will be fully briefed at all experimental trials, and demonstrations of the exercises required during testing will be conducted by the lead researcher to ensure clarity. All the risks associated with the proposed research have been identified along with appropriate preventative measures and are contained within relevant departmental risk assessment documentation. Copies of the risk assessments are available on request. All the elements of the research have been submitted for ethical approval in accordance with Edge Hill University guidelines, and your testing session will only commence once ethical approval has been granted. Your testing session will be conducted in the presence of a qualified first-aid trained member of staff. Please be reminded that your testing session will be terminated immediately should you display signs of discomfort/distress.

Safety

Health and safety procedures will be adhered to at all times, in line with the departmental health and safety manual. A thorough pre-testing questionnaire will determine your eligibility for the exercises involved in the research. All equipment required for the data collection process will be safety checked prior to use to minimise the risk to participants. A fully qualified first-aider will be on hand at all times throughout your testing session.

Benefits of Participation

Whilst the benefits of participation may not be obvious, there are benefits to your participation in the research nonetheless. You will gain experience of how research is conducted from inception to completion. Specifically, you will experience how a combination of sports science methods enable a detailed analyses of the mechanics of dance performance. You may develop an understanding of the mechanisms associated with injury risk in dance, which may be used to inform your future training practices. You will have to opportunity to ask questions relating to the research which may be an area of personal interest as a dancer. Finally, findings from the research may add to the body of knowledge within the dance science research domain, with the aim of minimising injuries to enhance performance.

Can you withdraw from the Research?

You can change your mind about participating in the research at any time during completion your experimental trials. You do not have to provide a reason, and there will be no consequences associated with your withdrawal. You have the right to withdraw your data within 4 weeks of the completion of your testing session, and can do so by contacting the lead researcher.

Confidentiality and Data Handling

You and your data will be anonymised by assigning you a unique numerical identification code. This code will be used for all recorded data to ensure your anonymity is maintained. Information relating to you such as your name, date of birth and contact details, will be stored on a password protected computer and back up via an encrypted external hard drive in accordance with the Data Protection Act 2003. All hard copy information enclosed within your informed consent sheet and your health screening forms, will be stored by the lead researcher in a locked filing cabinet within a secure office in accord with the Data Protection Act 1998. Only the research team will have access to your data. Following completion of the research project, your personal information will be destroyed and raw anonymised data will be retained for 10 years, in line with Edge Hill University's data policy. Your data may be used in peer reviewed publications or conference presentations, whilst preserving full anonymity.

Mistreatment

If at any stage of the research process you feel you have been mistreated or misinformed in anyway, you should contact the Departmental Ethics Committee Chair using the following contact details:

Professor Lars McNaughton – 01695587296. Email – Lars.mcnaughton@edgehill.ac.uk.

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Philip Nagy (Principal Investigator); Nagyp@edgehill.ac.uk - 07751619481

Dr. Matt Greig (co-principal investigator); Greigm@edgehill.ac.uk

Dr. Chris Brogden (co-principal investigator); Brogdenc@edgehill.ac.uk

Thank you for considering taking part in this research and sparing time to read the information provided. If you have any queries or would like more information, please don't hesitate to get in contact using the above details.

<u>Participant Informed Consent Form</u>

Title of Projects:

1) **The Kinematic and Kinetic Responses to Ballet-specific Movements**
2) **Electromyographic Responses to a Ballet-specific Movement Battery**
3) **Isokinetic Strength Profiling of Amateur Female Ballet Dancers**

Lay Title:

Biomechanical analysis of dance-specific tasks

Name of Researchers:

Philip Nagy: Nagyp@edgehill.ac.uk

Dr. Matt Greig: Greigm@edgehill.ac.uk

Dr. Chris Brogden: Brogdenc@edgehill.ac.uk

Please cross out which ever does not apply

Have you read the participant information sheet?

YES/NO

Have you had the opportunity to ask questions and discuss this study?

YES/NO

Have you received satisfactory answers to ALL of your questions?

YES/NO

Have you received enough information about the study?

YES/NO

Have you been made aware of any possible risks and discomforts?

YES/NO

Will you inform the researcher immediately if you are in pain or feel uncomfortable?

YES/NO

Do you understand you are free to withdraw from the study?

- At any time during the experimental trials, and up to four weeks after your final data collection session?

  YES/NO

- Without having to provide a reason?

  YES/NO

I agree to take part in the study.

Signed (Subject) **Date:**

Print (Subject)

Signed (Investigator) **Date:**

Print (investigator)

Project Title:

Within- and Between-day Load Responses to Ballet Training and Performance

Lay Title:

A GPS Analysis of Dance Performance

You are invited to take part in a research study within the Sport and Physical Activity Department at Edge Hill University. Before you decide whether to partake, it is crucial you understand why the research is being conducted, and, what will be required from you. Please allow yourself time to read the following document carefully and discuss it with friends, family and your GP if you wish. Do not hesitate to contact me if any of the elements in the research require clarity, or simply, that you would like more information.

Purpose of the Study

Previous research has identified high injury rates in dancers, with many injuries linked to repeated exposure to the demands of performance. Whilst there is an understanding of the types of injuries sustained in dance, there is little evidence regarding the influence of exposure to dance on injury risk. Therefore, using Global Positioning System (GPS) technology, the current research aims to investigate the demands of dance performance with implications on the risk of injury to dance performers.

Eligibility

You are invited to participate in the research given your status as a currently active dancer, and, your familiarity and experience with the movements associated with dance performance. You will be injury free in the six months leading up to your testing session. Participation is entirely voluntary and you will be asked to provide written informed consent should you wish to partake in the research. Participation reserves your right to withdraw from the research at any time during the experimental trials, and up to four weeks following your final data collection session without reason or consequence.

To determine your eligibility for the exercises in the study, you will undergo a thorough health assessment. The process will involve you completing an extended questionnaire to provide information regarding any current or previous medical issues, and a history of injuries. Should you present any medical or injury cases, you will be unable to take part in the testing procedures until further measures are applied, and in some instances, advised to seek the help of a medical practitioner. If you display, or inform the researcher of any signs of physical or mental distress during the data collection process, the session will be terminated.

Study Procedures

If you consent to take part, you will be asked to attend the Dance Studio at Edge Hill University (Wilson Centre, Department for Sport and Physical Activity), on three occasions. The data collection session will be arranged for a time and date that best suits you. The testing session will consist of familiarisation with the exercise and the experimental trial.

The entire data collection process is anticipated to last no longer than 2 hours, and a 24hr rest period will separate each testing session.

The familiarisation process will ensure you are clear and comfortable with the equipment and exercises required during the testing session. Details of the experimental components to which you will be familiarised with, are contained in the following sections. The experimental trail will comprise the completion of a submaximal multi-stage dance-specific choreograph comprised of various movements including jump-landing manoeuvres, with a graded increment in intensity between stages. A number of measures - as described in the subsequent section - will be recorded during the exercise. Prior to each of the experimental trails, you will be required to complete a 10 minute warm-up comprised of general mobilisation exercise and dynamic stretching.

Global Positioning System (GPS)

During the dance-specific choreograph you will be fitted with a neoprene vest containing a small matchbox-sized device, which will sit between your shoulder blades. The GPS unit will record continuously throughout each trial and will be used to measure 3D accelerations as a marker of exercise 'load'. The equipment should not provide you with any discomfort or compromise movement, and therefore, the risk to you is minimal.

Heart Rate

Completion of the multi-stage dance-specific choreography will require you to wear a heart rate monitor for the purposes of data collection. Your heart rate will be recorded throughout the exercise. As with the GPS unit, the heart rate monitor will not cause discomfort or restrict natural movement.

Rating of Perceived Exertion (RPE)

At the end of each stage of the protocol, you will be asked to rate how intense you perceive the exercise to be. The scores range from 6 (no exertion) to 20 (maximal exertion. A 6-20 chart with corresponding exertion descriptors will be provided to inform your decision. This is a personal reflection of exercise intensity.

Other Requirements

- You must be currently active in dance, and free from injury in the six months leading up to your testing session.
- For the purposes of data collection, you will be required to wear clothing similar to that worn during typical dance training, and standardized competition-specific footwear (i.e. ballet shoes). Please be advised that appropriate facilities will be provided for any changing requirements.
- You will be required to refrain from alcohol and vigorous exercise 24 hours prior to each experimental trial.

Risk to Participants

The procedures outlined in the experimental trial above are commonplace in sport and physical activity research. The testing exercises will be within the limits of those you experience during typical dance activity and therefore, the risk of injury is suggested to be minimal. All potential injury risks will be mitigated through medical and pre-exercise screening assessments, and you will be monitored closely for signs of pain and discomfort during completion of the exercise. You will be fully briefed before the experimental trial, and verbal instructions of the exercises required during testing will be provided by the lead researcher to ensure clarity. All the risks associated with the proposed research have been identified along with appropriate preventative measures and are contained within relevant departmental risk assessment documentation. Copies of the risk assessments are available on request. All the elements of the research have been submitted for ethical approval in accordance with Edge Hill University guidelines, and your testing session will only commence once ethical approval has been granted. Your testing session will be conducted in the presence of a qualified first-aid trained member of staff. Please be reminded that your testing session will be terminated immediately should you display signs of discomfort/distress.

Safety

Health and safety procedures will be adhered to at all times, in line with the departmental health and safety manual. A thorough pre-testing questionnaire will determine your eligibility for the exercises involved in the research. All equipment required for the data collection process will be safety checked prior to use to minimise the risk to participants. A fully qualified first-aider will be on hand at all times throughout your testing session.

Benefits of Participation

Whilst the benefits of participation may not be obvious, there are benefits to your participation in the research nonetheless. You will gain experience of how research is conducted from inception to completion. Specifically, you will experience how a combination of sports science methods enable a detailed analysis of the mechanics of dance performance You may develop an understanding of the mechanisms associated with injury risk in dance, which may be used to inform your future training practices. You will have to opportunity to ask questions relating to the research which may be an area of personal interest as a dancer. Finally, findings from the research may add to the body of knowledge within the dance science research domain, with the aim of minimising injuries to enhance performance.

Can you withdraw from the Research?

You can change your mind about participating in the research at any time during completion of your experimental trials. You do not have to provide a reason, and there will be no consequences associated with your withdrawal. You have the right to withdraw your data within 4 weeks of the completion of your testing sessions, and can do so by contacting the lead researcher.

Confidentiality and Data Handling

You and your data will be anonymised by assigning you a unique numerical identification code. This code will be used for all recorded data to ensure your anonymity is maintained. Information relating to you such as your name, date of birth and contact details, will be stored on a password protected computer and back up via an encrypted external hard drive in accordance with the Data Protection Act 2003. All hard copy information enclosed within your informed consent sheet and your health screening forms, will be stored by the lead researcher in a locked filing cabinet within a secure office in accord with the Data Protection Act 1998. Only the research team will have access to your data. Following completion of the research project, your personal information will be destroyed and raw anonymised data will be retained for 10 years, in line with Edge Hill University's data policy. Your data may be used in peer reviewed publications or conference presentations, whilst preserving full anonymity.

Mistreatment

If at any stage of the research process you feel you have been mistreated or misinformed in anyway, you should contact the Departmental Ethics Committee Chair using the following contact details:

Professor Lars McNaughton – 01695587296. Email – Lars.mcnaughton@edgehill.ac.uk.

Researcher Contact Details

Philip Nagy (Principal Investigator); Nagyp@edgehill.ac.uk - 07751619481

Dr. Matt Greig (co-principal investigator); Greigm@edgehill.ac.uk

Dr. Chris Brogden (co-principal investigator); Brogdenc@edgehill.ac.uk

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Print (Subject)

Signed (Investigator) **Date:**

Print (investigator)