UNIVERSIDADE DE LISBOA

Faculdade de Motricidade Humana

Dancers' Landing Strategies – Experimental and Computational Approaches

Ana Paula Morais Teixeira de Azevedo

Orientador: Professor Doutor Raul Alexandre Nunes da Silva Oliveira
Co-orientadores: Professor Doutor Nelson Filipe Neves Cortes
Professor Doutor João Pedro Casaca Rocha Vaz

Tese especialmente elaborada para obtenção do grau de Doutor no ramo Motricidade Humana, na especialidade de Comportamento Motor

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Faculdade de Motricidade Humana – Universidade de Lisboa
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Assinatura:____
(Ana Paula Morais Teixeira de Azevedo)

DEDICATÓRIA

Para a minha irmã!

GRANTS, PEER-REVIEWED PAPERS & INTERNATIONAL CONFERENCE ABSTRACTS, AND AWARDS DURING THE PhD

During the PhD program, two outstanding visiting research scholarships were awarded to further develop and improve research skills. Additionally, the PhD findings have been disseminated (2018) and are in the process of being presented (2019) at international conferences or annual meetings. The scientific meetings focus on fields related to this PhD thesis, such as Biomechanics, Dance Science, Bioengineering, and Computational Modeling.

A. Grants

Academic year 2017-2018:

- **Fulbright Visiting Research Scholarship** at George Mason University (VA, USA); Mentor Dr. Nelson Cortes (Associate Professor). Duration: 9 months.

Academic year 2018-2019:

- **Short-term Scholar – Visiting Researcher** at George Mason University (VA, USA); Mentor Dr. Nelson Cortes (Associate Professor). Duration: 6 months.

B. Peer-reviewed Papers

2019

- **Published:** Azevedo AM, Oliveira R, Vaz JR, & Cortes N (2019). Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings. *Med Sc Sport Exerc*, 51(3): 539-547.

- **Published:** Azevedo AM, Oliveira R, Vaz JR, & Cortes N (2019). Foot Modeling Affects Ankle Sagittal Plane Kinematics during Jump-landing. *J Biomech*, https://doi.org/10.1016/j.jbiomech.2019.109337

- **Accepted** – to be **published** in indexed *IEEE Xplore and Medline/PubMed*: Azevedo AM, Oliveira R, Vaz JR, & Cortes N (2019). Effect of Two Different Pose

Estimation Approaches on Lower Extremity Biomechanics in Professional Dancers, *IN PRESS.*

C. Peer-reviewed Conferences or Annual Meetings

2018

- June – **Thematic Poster**

  65th American College of Sports Medicine Annual Meeting (Minneapolis, MN, USA). **The Biomechanical Pattern of Multidirectional Single-leg Landing in Professional Dancers and Non-dancers.**

- August – **Poster Presentation**

  42nd American Society of Biomechanics Conference (Rochester, MN, USA). **Landing Biomechanics differ between Professional Dancers and Non-dancers: Experimental and Preliminary Modeling Approaches.**

- September – **Oral Presentation**

  2nd Lisbon Foot & Ankle Clinical Biomechanics Course 2018, Sports Edition (Lisbon, Portugal). **Movement Analysis in Dancers – Landing Biomechanics**.

- October – **Oral Presentation**

  28th Annual Meeting of the International Association for Dance Medicine & Science (Helsinki, Finland). **Professional Dancers have Distinct Multi-segmented Foot-ankle Biomechanics than Non-dancers during Multidirectional Landings** (2018 IADMS Research Student Award).

2019

- July – **Ignite Session and Poster Presentation**

  41st Engineering in Medicine and Biology Conference (Berlin, Germany). **Effect of Two Different Pose Estimation Approaches on Lower Extremity Biomechanics in Professional Dancers**.

- August – **Poster Presentation**
  27th Congress International Society of Biomechanics and 43rd American Society of Biomechanics Conference (Calgary, Canada). **Musculoskeletal Computational Model Optimization: The Critical Role of Markers Error Adjustments**.

- October – **Poster Presentation**
  29th Annual Meeting of the International Association for Dance Medicine & Science (Montreal, Canada). **Professional Dancers Shock Absorption Mechanism Differs from Non-dancers during Landings**.

D. Conference Award

2018

The oral communication **Professional Dancers have Distinct Multi-segmented Foot-ankle Biomechanics than Non-dancers during Multidirectional Landings**, which is part of this PhD thesis and aforementioned, was distinguished with the Conference Award, **2018 IADMS Research Student Award** at the 28th Annual Meeting of the International Association for Dance Medicine & Science (Helsinki, Finland).

TITTLE: Dancers' Landing Strategies – Experimental and Computational Approaches

ABSTRACT

Jump-landing activities are an essential part of dance training technique, and have aesthetic and technic constrains. Previous research reported that dancers display an extended landing posture at initial ground contact, which has been considered a deleterious pattern. However, this population shows higher joint excursions of the lower extremity allowing to appropriately dissipate the impact forces. Additionally, computational modeling and simulation (e.g., OpenSim) has had limited applications in dance science. This PhD thesis main purpose was to determine lower extremity biomechanical patterns in professional dancers during single-leg landings in lateral, diagonal and forward directions and compare to non-dancers. Five studies were conducted employing experimental and computational modeling approaches: 1) a more broad study that exhibited extensive kinematics and kinetic differences between professional dancers and non-dancers in multidirectional jump-landings; 2) a study that explored the different foot modeling and highlighted lower ankle magnitudes in the multi-segmented foot model than the single rigid foot model; 3) an investigation that revealed higher kinematic magnitudes in professional dancers foot-ankle complex, providing additional data related to foot kinematics (e.g., forefoot-hindfoot); 4) an experiment that ensured that knee and ankle sagittal plane kinematics were less sensitive to the kinematic model chosen than hip joint, when comparing OpenSim and Visual 3D methods; 5) a modeling approach that sought to determine the impact of different processing phases while developing a kinematic model on the sagittal lower extremity kinematics. Overall, the findings of this PhD provided scientific data for dance science, supporting that some characteristics of professional dancers landing patterns may be appropriate for energy dissipation and therefore may contribute to the shock absorption mechanism in jump-landings. Moreover, it is important to recognize that OpenSim is a multistep process, and each stage is interdependent with potential interlinked effects, where the user responsibility is to ensure that the model represents the experimental data and the simulation is reliable. Further research is needed to decrease such subjectivity in the modeling approach. Additional investigation is also required to continue exploring jump-landings to support risk management and development of effective intervention and prevention programs.

KEYWORDS: Biomechanics; Dance Science; Single-leg landing; Multi-segmented foot model; OpenSim.

TÍTULO: Análise Biomecânica do *Landing* em Bailarinos Profissionais.

RESUMO

As actividades de *jump-landing* são parte essencial no treino dos bailarinos. Estes apresentam uma postura mais em extensão no instante do contacto inicial com o solo, após um salto. Este padrão é associado a uma menor eficiência na dissipação das forças de impacto. Contudo, os bailarinos apresentam grandes amplitudes de movimento nas articulações do membro inferior, o que poderá permitir uma absorção das forças de impacto mais eficaz. Por outro lado, investigação na área da dança tem explorado pouco os modelos e simulações computacionais (ex. *OpenSim*). Esta tese de Doutoramento teve como principal objectivo investigar o padrão biomecânico do membro inferior dos bailarinos profissionais, nas actividades de *jump-landing* em várias direcções, em apoio unipodal. Ao longo deste projecto de investigação, cinco estudos foram realizados utilizando dois métodos, experimental e computacional: 1) estudo mais abrangente que evidenciou as diferenças cinemáticas e cinéticas entre bailarinos profissionais e não-bailarinos, durante *jump-landings* multidirecionais; 2) estudo que explorou o *modeling* do pé e tornozelo, revelando que o modelo (segmento rígido vs. multissegmentado) afecta a análise cinemática do tornozelo (menores valores no modelo multissegmentado); 3) estudo que corroborou que os bailarinos profissionais apresentam maiores amplitudes articulares no complexo articular pé-tornozelo comparado aos não-bailarinos, o que contribuiu para o conhecimento mais real sobre a cinemática e participação dos vários segmentos do pé (ex.: retro-pé, médio-pé); 4) estudo onde se verificou que os ângulos articulares da anca, no plano sagital, são mais sensíveis ao modelo cinemático escolhido quando se utilizam dois *software* diferentes (*OpenSim* e *Visual 3D)*; 5) estudo que demonstrou como os ângulos articulares do membro inferior são afectados em diferentes fases de processamento, aquando do desenvolvimento de um modelo cinemático. Os resultados deste projecto de Doutoramento providenciaram informação relevante para a *dance science*, enfatizando que o padrão de *landing* dos bailarinos profissionais pode trazer vantagens para a dissipação de energia e consequentemente poderá contribuir para um mecanismo de absorção do impacto mais eficiente. No *OpenSim*, o investigador é responsável por assegurar que o modelo representa os dados recolhidos. Estudos são necessários para minimizar a subjetividade inerente ao processo de desenvolvimento de modelos e simulações computacionais. Este software apresenta potencial para se investigar actividades de *jump-landing* e assim poder contribuir para melhorar a gestão de risco e fundamentar programas de intervenção e prevenção de lesões.

PALAVRAS-CHAVE: Biomecânica; *Dance Science*; *Landing* em apoio unipodal; Modelo do pé multissegmentado; *OpenSim*.

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LIST OF ABBREVIATIONS

DJ	Diagonal Jump-landing
DOF	Degrees of Freedom
FJ	Forward Jump-landing
IC	Initial Contact
LE	Lower Extremity
LJ	Lateral Jump-landing
Max	Maximum
PKF	Peak Knee Flexion
PvGRF	Peak vertical Ground Reaction Force
RMS	Root Mean Square
SD	Standard Deviation

CHAPTER I

1. Introduction

Dance movements result from a set of coordinated biomechanical degrees of freedom, ruled by technical and aesthetic requirements (1). The development of dance aptitudes requests continuous and progressive work, beginning at an early age and undertaken over years (2, 3). Professional dancers represent the elite that achieve the higher levels of the profession (2). In dance, as in other athletic activities, the injury rates have been a hot topic, resulting in several studies. In these two populations, the lower extremity has an expressively higher number of musculoskeletal injuries compared to other anatomical structures (4-10). Interestingly, it has been reported that acute lower extremity injury rate is lower in dancers, compared to other athletic populations (4-6, 10). Furthermore, dancers are frequently exposed to jump-landings that are usually performed with horizontal displacement of the body with single- or double-leg support, in many directions, as part of their training technique to use a wide range of the stage or classroom. Jump-landing proficiency is thereby a request for dancers to progress in the professional career (11). They are not only exposed to the strenuous physical demands placed on their bodies, but also to the consequences of repetitive jump-landing performances, constrained by technical and aesthetic requirements (12, 13).

Several studies have examined jump-landings, predominantly using drop landing tasks, to propose strategies to target biomechanical modifiable risk factors (13-21). Extended landing posture is, for instance, one of the examples. It has been associated with less ability of the lower extremity to mitigate the landings forces, and consequently produces a higher peak vertical ground reaction force (PvGRF) (17, 18, 22). Despite many studies proposing injury prevention programs, the injury rates remain steady across populations and years (7-9, 23), which undoubtedly is a key-problem in athletic activities. Based on previous research, dancers display an extended landing posture at initial contact (IC) of the landing phase (13, 15, 24). Yet, most likely this should not be seen as a deleterious factor, since dancers are encouraged to use more lower extremity joint excursions (15), particularly of the knee and foot-ankle joints, with longer landing phase that may suggest impact forces been absorbed over a longer period of time. Injury risk factors still need a thorough biomechanical investigation to develop and implement effective intervention and injury prevention programs.

Motion analysis in dancers and the effect of dance training on dancers' body is evolving. Still, few research has been examining dancers jump-landings in multiple directions with a single-leg support. The landing phase of a jump-landing is a demanding task that involves rapid deceleration to "brake" the movement, and where considerable amount of forces are acting on the lower extremity. This becomes more challenging when the base of support is just one limb (25, 26), and the motion is taken in different directions (27), which have a great effect on the biomechanical profiles. Also, research incorporating a multi-segmented foot model during landing is required to better understand the intricate structure of the foot and ankle joints, particularly due to its influence on other lower extremity joints during landing (19, 21, 28).

Additionally, computational modeling and simulation (e.g., OpenSim) is an impressive tool to study variables difficult to obtain experimentally (e.g., muscle and joint contact forces) (29-32). It is a field that has not been yet sufficiently explored in dance science, and can be valuable to further study the mechanisms underlying dance movement, particularly lower extremity joint functions during jump-landings. However, despite the considerable value to gain insights into human movement using OpenSim, it is a complex and multistep process (31, 33). The implementation of this approach not only involves a subjective process, but also is dependent on users' skills and experience (34, 35). During that process, manual adjustments of the markers are needed to achieve effective marker set optimization. Therefore, the development of a musculoskeletal model can produce errors that can propagate through the workflow, affecting interpretation of the outputs (34, 36).

Hence, in this PhD research, an experimental and computational modeling and simulation approaches were conducted. The purpose was to investigate whether professional dancers landing pattern would possibly exhibited alternative strategies to implement in future intervention and injury prevention programs. The leading focus of this PhD research involved the study of the lower extremity biomechanical patterns of professional dancers and non-dancers, while performing multidirectional single-leg landings. For that, 3D motion analysis, force plate, and computational modeling and simulation were used. Accordingly, this PhD research sought to contribute not only to dance science, but also to provide further evidence-based information regarding jump-landing performances, that might be beneficial for sports sciences to potentially contribute to decrease the risk of injury rate in jump-landings.

Rationale, Research Questions, and Hypotheses

The main purpose of this PhD thesis was to investigate professional dancers' lower extremity biomechanical patterns during single-leg multidirectional jump-landings. To accomplish this aim, a scientific research process was conducted employing a single-session descriptive group comparison design. Each study had its specific research question and respective hypothesis based on a scientific rationale. This is presented through the five scientific papers that have been published, accepted or are currently under review to international peer-reviewed journals. Paper 5 (Chapter VII) is still in preparation to be submitted in a near future.

- **Paper 1**

Title: *Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings.*
Status: Published.

Rationale: It is well reported that dancers have a significantly high overuse injury rate, mainly on the lower extremity. Conversely, the lower extremity acute injury rate is considerably lower in dancers when compared to other athletic populations. Previous research has proposed optimal jump-landing biomechanical patterns to attenuate the deleterious factors during landing (e.g. high impact forces). It is also established that injury prevention programs have targeted modifiable risk factors to decrease injury rates in jump-landing activities. Still, overall lower extremity injury rates in many athletic populations have remained steady over the past decade. Dance, as other athletic activities, is characterized by a significant volume of jump-landing tasks. Thus, dancers are exposed to jump-landing training technique since young age, with required aesthetic and technique constraints. Due to insufficient scientific evidence related to dancers single-leg multidirectional landings, this study presented the findings of the lower extremity biomechanical landing patterns in professional dancers and respective comparison to non-dancers'. Additionally, this study can provide further understanding of the importance of the jump-landing directions to implement in prevention and intervention programs.

Research question: Are there differences in the lower extremity biomechanical parameters between professional dancers and non-dancers during single-leg multidirectional jump-landings?

Hypothesis:

i) Professional dancers would demonstrate a more extended landing posture (lower hip and knee flexion, and higher plantarflexion), higher ankle excursion, and PvGRF during single-leg jump-landings in multidirectional jump-landing activities, compared to non-dancers.

ii) Lateral jump-landing direction would have higher knee and lower hip flexion angles than the other two directions (forward and diagonal).

- **Paper 2**

Title: *Foot Modeling Affects Ankle Sagittal Plane Kinematics during Jump-landing.*
Status: Published.

Rationale: Based on the findings of paper 1, a natural progression was to focus on the foot-ankle complex. It is a noteworthy anatomical structure that initially attenuate the impact forces and is of utmost importance in dancers. It is a key-element in jump-landing activities and also has influence on the other joints of the lower extremity. Considerable biomechanical studies have employed the single-segmented foot to analyze this intricate structure. However, differences in the ankle kinematics have been reported when using the multi- or the single-segmented foot models, which may lead to inaccuracies on ankle joint kinematics. Furthermore, the single foot model does not consider the other foot joints. Still, there are limited studies investigating ankle kinematics between multi- and single-segmented foot models during single-leg multidirectional jump-landings. Thus, this study compared ankle kinematics between the two foot models (multi-segmented and single-segmented) in jump-landing tasks.

Research question: Are there differences in ankle joint kinematics computed by a multi-segmented foot model (Oxford Foot Model) and a single-segmented foot model (conventional ankle) in multidirectional single-leg landings?

Hypothesis:
The hindfoot-tibia angles and excursion magnitudes would be lower in the multi-segmented foot model compared to the ankle of the single-segmented foot model.

- **Paper 3**

Title: *Oxford Foot Model Kinematics in Landings: a Comparison between Professional Dancers and Non-dancers.*
Status: Under second revisions.

Rationale: When quantifying the ankle kinematics, in paper 2, differences were found between the multi- and single-segmented foot models. The foot-ankle complex has a remarkable importance in jump-landing tasks; it allows to initially mitigate the landing forces, to support the body, and has an influence on the other lower extremity joints. Dancers not only often perform jump-landings, but also commonly rely on this anatomical structure while dancing. Most biomechanical studies in dancers jump-landings have been using the single foot model. However, the foot-ankle complex is an intricate and frequently required structure in dance; and limited research exists regarding dancers multi-segmented foot model during single-leg jump-landings. Hence, this study took into consideration the different foot segments (hindfoot, forefoot, hallux), using the Oxford Foot Model, and compared dancers' and non-dancers' kinematics of the foot-ankle complex.

Research question: Are there differences in the multi-segmented foot (Oxford Foot Model) kinematics between professional dancers and non-dancers, during lateral and forward single-leg landings?

Hypothesis:
Professional dancers would have higher plantarflexion and eversion angles and excursions in the foot and ankle joints, using the multi-segmented foot model, than non-dancers.

- **Paper 4**

Title: *Effect of Two Different Pose Estimation Approaches on Lower Extremity Biomechanics in Professional Dancers.*
Status: Accepted for publication.

Rationale – Musculoskeletal modeling and simulation of dynamic activities permit the analysis and estimation of athletic performance. Different modeling approaches have been used to explore lower extremity biomechanics in jump-landing tasks. While computational modeling has been employed in many fields, it is significantly lacking in the dance science field. Visual 3D and OpenSim are commonly used software in biomechanical studies. The main difference between these two approaches is the algorithm used to estimate the position and orientation of musculoskeletal models. Visual 3D uses segment optimization, whereas OpenSim uses global optimization. For both approaches, pros and cons have been reported in the literature. During the earlier phases of the PhD work, Visual 3D was the software chosen to compute joint angles, and moments. However, to further explore and apply computational modeling into this project, OpenSim software was also selected with the purpose to conduct inverse kinematics, inverse dynamics, muscle analysis, and to compute lower extremity joint loads. The differences in these two approaches' algorithms merit further investigation. Thus, this preliminary study was pertinent to understand whether the different approaches were providing similar outputs of the studied variables, using the same data set of a single-leg forward jump-landing. This comparison is considerably important for the next steps, after the PhD, concerning muscle analyses and joint reaction forces estimations on the lower extremity joints.

Research question – Do different pose estimation approaches (Visual 3D and OpenSim) have an effect on the hip, knee, ankle joint angles and moments in professional dancers, during a single-leg forward jump-landing?

- **Paper 5**

Title: *The Impact of Marker Adjustment in Musculoskeletal Modeling on Kinematic Parameters.*
Status: In preparation to be submitted.

Rationale – The accuracy level of a musculoskeletal model depends on the application and not only on the model itself. During the process of biomechanical studies to create a musculoskeletal model, errors in early stages of the model development, such as marker errors, can propagate and affect the results leading to biomechanical inaccuracies. Currently, in computational modeling using OpenSim software, it is essential to compute a musculoskeletal model. However, it is a subjective process that relies on the user skills and experience. In inverse kinematics, for effective marker set optimization it is essential to identify, examine and adjust all markers with max error above the recommended threshold. This is important to ensure that the model is adequately representing the musculoskeletal system that is simulating. Understanding the importance of this step is critical, since a lack of emphasis has been given on the sensitivity of the models and dependency on user skills, which may influence the final biomechanical outputs.

Research question – Are there differences in the sagittal lower extremity joint angles in three difference processing phases of developing a musculoskeletal model i) before markers adjustment, ii) after single max marker error adjustment, and iii) after all makers with max errors adjusted?

Thesis Structure

The PhD thesis structure is organized in X chapters, assembling information needed to understand the conducted research:

- **Chapter I – Introduction**: introduces the main research topics of the PhD thesis, including the significance of the conducted research. It also displays the rationale behind all the research done, as well as, the research questions, and hypotheses for each study;
- **Chapter II – Literature Review**: summarizes the state of the art of the main research topics of the PhD thesis. It contributes to address different and fundamental topics concerning research in multidirectional single-leg jump-landings, such as dance, jump-landing biomechanics, foot-ankle complex and computational modeling and simulation.
- **Chapter III to VII – Research Studies**: presents the five scientific studies written during the PhD process. Each of them was organized following the structure of a scientific paper format: Introduction, Methods, Results, Discussion, Conclusions, and References; and each one was reformatted for consistency with the thesis. The original paper already published can be seen in Appendix.
- **Chapters VIII to IX – General Discussion and Conclusions**: provides the main findings of chapters III to VII, and the overall conclusion of this PhD thesis. Additionally, based on the main findings it also includes limitations, recommendations for future research, and practical implications.
- **Chapter X – List of References**: enumerates the used references in the chapters Introduction, Literature Review, General Discussion, and Conclusions. This chapter does not include the references of each study, since they are already included in their own chapter.

Chapter II

2. Literature Review

2.1 Dance

Dance is a demanding and elaborate form of body movement that requires development and proficiency of specific skills over manifold years. Due to the physical and aesthetic demands during dancers' careers, they acquire and develop singular movement patterns of the musculoskeletal system. This is also a result of the unique anatomical positions often performed, particularly noticeable in the lower extremities, leading dancers to a distinctive biomechanical pattern. Since a young age, they are exposed to high physical demands placed on their bodies while performing dance technique (5, 11, 13, 37). This daily occurrence can lead to a high risk of musculoskeletal injuries (4, 5, 38) that can have considerable repercussions throughout their lifespan. On the other hand, dancers combine technique with body alignment and postural control while dancing, increasing awareness of their body position and orientation compared to other athletic populations.

In an effort to better understand the underlying movement performance, numerous biomechanical studies have been conducted in athletes (7-9, 14, 17-20, 22, 23, 26, 27, 39-53). Though, in dance science, there is still a lack of biomechanical research. Due to the distinctive and singular technique, undertaken through many years of intensive training, more extensive studies are needed. Furthermore, it is well-established the importance of studying movement biomechanics to determine and assess injury risk factors. This can eventually contribute to determine strategies that may be implemented in intervention and injury prevention programs (54).

2.1.1 Dancers' injury epidemiology of the lower extremity

Dancers have a high injury rate with more than 60% of all injuries occurring to the lower extremity (4-6); which is similar to other athletic populations (7-10). In dance, the most common anatomical locations of injuries in the lower extremity are: the knee (7% to 15%) (4-6), lower leg (9% to 18%) (4, 5), ankle (14% to 33%) (4-6), and foot (9% to 20%) (4-6). Dancers' injuries are commonly linked to stress-related factors, involving tendons, ligaments and/or muscles (4). These injuries are frequently attributed to overtraining, repetitive dance movements and the demanding extreme ranges of motion. Consequently, high physical demands are placed on the lower extremity with excessive microtrauma in the anatomical structures (5, 38). Accordingly, 66% to 72% of the injuries in dancers are classified as overuse (4-6), resulting from cumulative microtrauma or repetitive use and stress (10, 55). Those injuries in dancers have a greater incidence and severity than traumatic injuries (5). Subsequently, the remaining 28% to 34% of the injuries have been reported as acute (4-6), which is about 2.5 times lower compared to other athletic populations – i.e., approximately 70% of the injuries are acute in nature (10).

Acute and overuse injuries often limit unrestricted athletic practice (23). The injuries can lead to short and long-term health consequences, affecting performance and quality of life with potential for extended disability (54) in dancers and athletes' lifespan. Due to their technique, dancers are exposed to similar overuse injuries as other athletic activities (56), such as tendinopathies or stress fractures (4, 55). Nonetheless, dancers have unique dance-related injuries (56), such as foot strain (5) or snapping hip (4). Additionally, there are movement patterns commonly seen in dancers' technique that are not observed in other athletic populations. Few examples are the knee flexion immediately before and after a jump (known as *demi-plié*) or rolling the feet in landing phase (12, 13, 57). Interestingly, these dance features have been associated to biomechanical improvements in jump-landing tasks, to reduce the risk of injuries. These dance-related characteristics will be further analyzed in Chapter II – section 2.2.1.

2.1.2 Modifiable risk factors related to the lower extremity

Lower extremity musculoskeletal injuries have a multifactorial etiology (58). Injury risk factors must be clearly established, so prevention and intervention programs can be devised to adequately and successfully target them (58). The risk factors interact in multiple ways. The modifiable risk factors can be subject to intervention by physical training or behavioral approaches (54). Due to an extensive high injury rate in the lower extremity, a particular emphasis should be given on this anatomical region concerning prevention and intervention programs (7). Deficits in muscle strength and/or muscle imbalance (54, 59), altered neuromuscular control (22, 42, 59, 60), extended landing posture (17, 18), decreased ankle excursion in landings (51, 61, 62), and poor technique (38, 58, 63) are some of the well-established modifiable risk factors commonly associated to lower extremity injuries. It is noticeable that in the past decades, considerable research has been conducted to identify those factors, particularly during jump-landing activities (13, 15-19, 22, 26, 27, 39, 41, 42, 48, 64). Still, tracking and managing risk factors in biomechanics research remains a challenge. Injury prevention programs in athletic populations have been shown to be effective in changing some of the proposed modifiable risk factors and decreasing injury rates in the specific study sample (49, 50, 65). Nevertheless, despite extensive research and development of prevention programs, lower extremity injury rates have remained steady across multiple populations and years (7-9, 23). Additionally, the implementation, adoption, and maintenance of these programs in real life remain a challenge. Injury prevention programs should involve wider, appropriate and engaging implementations to achieve its goals. Therefore, it is important to identify and understand the movement behind jump-landing activities to adequately target the modifiable musculoskeletal risk factors linked to the lower extremity injuries.

2.1.3 Dancers training technique

Professional dancers emerge as a result of a demanding selection process, representing the elite of dance hierarchy (2, 3). Those dancers undertake a rigorous dance training over numerous years. High levels of performance in dance require several specific athletic skills, such as muscular strength and endurance, anaerobic and aerobic energy consumption, speed, agility, coordination, motor control and psychological readiness (63). The development of those skills requires a continuous and progressive work and is critical to dancer's performance. Moreover, for functional and aesthetics purposes, the technique requirements of dance performance not only involve considerable amount of *en dehors* (external rotation of the lower extremities), joint mobility, and muscle flexibility and strength, but also within extreme ranges of motion in several joints. Since early ages of dance training there is a strong emphasis on body alignment, particularly in *ballet* dancers. These dancers are trained to be in an upright upper body and neutral pelvis position. Additionally, both lower extremities are externally rotated (*en dehors*), while keeping the patella aligned with the second metatarsal. It is further expected to have neutral position of the hindfoot, associated with strong feet arches (2, 12, 13). Subsequently, all the aforementioned attributes are considerably important in training technique, making dancers a unique population, with distinctive biomechanical pattern compared to other athletic populations. Some of the dance training effects have been reported in previous literature, particularly in jump-landing activities.

Dancers are characterized by a smooth landing sequence, a request in their technique, for the audience to not visualize the movement effort (11). Also, they are encouraged in their training process to initially contact the ground with the toes in a fully plantarflexed position, followed by lowering through the foot, until a controlled heel contact (13, 24, 57). This increases the ankle range of motion, the peak ankle dorsiflexion and the knee flexion angles during landing, producing a deeper *demi-plié*, while controlling eccentric loads (13, 24). These dance-related features will be further examined in Chapter II – section 2.2.1. Hence, the lower injury rate of severe traumas (e.g., anterior cruciate ligament injury) in dancers when compared to athletes (11), may be explained by the combination of repetitive and specialized training that dancers undergo to obtain jump-landing and balance skills, along with longer time to achieve fatigue (15, 16, 24). Further, the relatively predictability nature of dance environment, which does not involve ball-handling as in other athletic populations (11), may contribute to the lower injury rate. Thus, observational and measurement of performance variables in dance through biomechanical analysis is important to enhance the understanding and provide thorough information of dance movements and skills in jump-landing activities.

2.2 Jump-landing biomechanics

For the development of prevention and intervention strategies, it is imperative to identify and understand injury mechanisms (45, 54). Jump-landing tasks are fundamental features of many athletic and dance activities and have received substantial research attention in the last decades (11, 13-16, 19, 20, 22, 24, 26, 27, 39, 40, 42-44, 47, 53, 66). The study of jump-landing biomechanics allows to investigate the demands placed on the body while performing these tasks (66). Jump-landing sequence can be divided into two phases: i) the flight phase in which the foot is off the ground, and ii) the stance phase when it is contacting the ground. Additionally, the stance phase (Figure 1) can be divided into: a) landing phase, which is corresponding to the deceleration phase, defined as the time period from IC with the ground to the peak knee flexion (PKF) (13, 51); and b) take-off phase, defined as the time period between the PKF and the toe off, when the foot stop contact with the ground. In this PhD research the prime focus was in the landing phase. It corresponds to the period of the greatest loading; when considerable large forces are applied to the human body (14, 41, 52, 65).

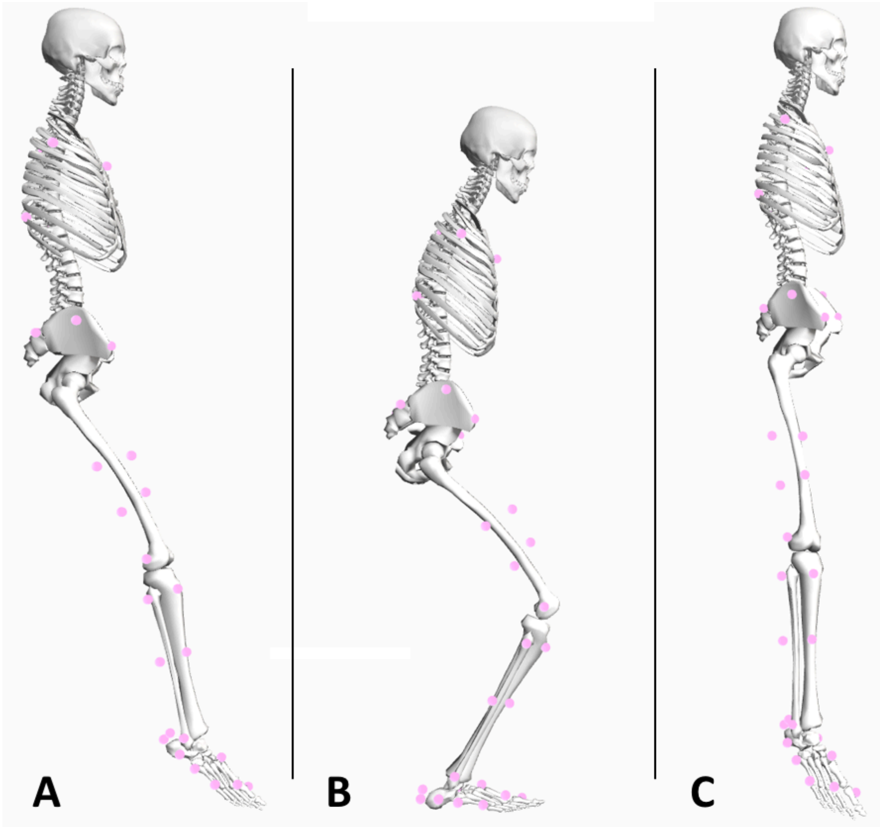

Figure 1. Sagittal view of the stance phase of a jump-landing: (A) initial contact; (B) Peak knee flexion; (C) and toe off events; (A) to (B) represents the landing phase, and (B) to (C) represents the take-off phase.

The lower extremity absorbs the body's energy during landing (53). A primary goal in this phase is to stabilize the body and maintain the center of mass within the base of support. For that, the lower extremity plays an important role, to reduce and control the downward momentum acquired during the flight phase; and also, to dissipate that kinetic energy generated (19, 44, 67). Altogether, it places strenuous demands on the lower extremity (67).

During jump-landings, injuries may occur from a multifactorial etiology, including repetitive malalignment of the lower extremity segments or from inadequate absorption of impact forces (61). A more flexed landing pattern, with higher trunk, hip and knee flexion during the landing phase (Figure 2) is associated with higher shock absorption mechanism, important contributor to energy dissipation, and consequently lower ground reaction forces (17, 18, 22). Thus, when employing an active trunk flexion strategy during landing, decreased PvGRF, peak knee extensor moment, higher hip and knee flexion angles and hip extensor moment have been reported (17, 18, 46). Though, despite the magnitudes of the attained hip and knee flexion (51º and 82º, respectively) being suggested to be protective of excessive joint loads (17, 18), it is plausible that such magnitudes are most likely incompatible with athletic performance, due to the concomitant excessive active trunk flexion (i.e., 96º of active trunk flexion), as depicted in Figure 2.

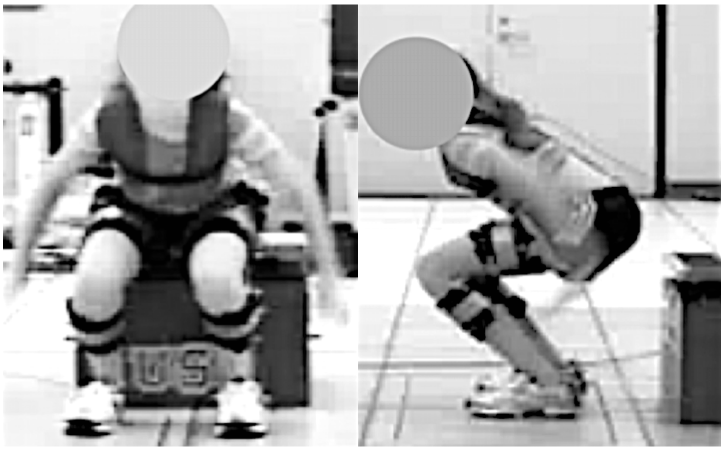

Figure 2. Example of an anterior (left) and sagittal (right) views of large trunk flexion strategy during double-leg drop landing task (65).

Being at a nearly fully extended position at IC it maximizes the range of motion (known as excursion in Chapters III to V) over which flexion can occur to dissipate energy (67). The lower extremity flexion during landing, controlled by extensor musculature eccentric work, contributes to the shock absorption mechanism (67). This is further beneficial as the increased sagittal joint excursions are related to concurrently reduced frontal plane joint excursions (14). There is also strong evidence suggesting that a decreased weightbearing ankle sagittal excursion in landing influences lower extremity mechanics and may be related to injuries due to a compensatory mechanical pattern in the lower extremity (37, 62). As an example, excessive foot pronation or compromised medial knee alignment (37). Additionally, the foot landing technique (e.g., toe-to-heel versus heel-to-toe landing) has noteworthy implications regarding the forces transmitted to the body and the body's ability to dissipate these forces (53, 67, 68). It has been suggested that a toe-to-heel landing technique leads to a considerably lower hip flexion angle at IC compared to the rearfoot (defined in this PhD thesis as hindfoot) and self-preferred landing techniques (53). Biomechanically, in the toe-to-heel strategy, a relatively large lever arm exists between the point of force application and the ankle joint, increasing the deceleration distance and thus decreasing the impact experienced (67). Increased plantarflexion angle at IC also provides a mechanical advantage, since it permits a subsequent increased ankle excursion during landing, established as an important contributor to shock absorption (19, 21, 41, 67). Accordingly, jump-landing technique influences the lower extremity ability to mitigate the impact forces. The landing technique therefore plays an important role, as it may influence the likelihood of injury (67) .

2.2.1 Jump-landing patterns in dancers

Biomechanical research is an emerging field in dance science. Dancers training places a significant focus on specific motor organization patterns to develop the performance and technique requirements (1). While substantial research studying the biomechanical responses of the lower extremity during landings exist, there is a paucity of these studies in dancers. Jump-landings are an important aspect of many dance styles (24). Accomplishment of aesthetically precise balance and jump-landing skills are necessary for dancers to advance in the professional careers (11). As in other athletic populations, dancers are often exposed to jump-landing activities; being part of their training technique throughout the years of practice (11, 69); a *ballet* dancer can perform up to 300 jumps per hour (69). Besides, dancers are not only exposed to the effects of repetitive jump-landings, but also have to be aware and develop the aesthetic requirements during their performances that other athletic populations do not have (11-13, 15, 70). Some of those aesthetic requirements are the alignment of the lower extremity (e.g., patella aligned with the second metatarsal) and the foot work (e.g. toe-to-heel landing) that subsequently allows extensive plantarflexion angle in the ankle joint (11, 13, 57). Hence, there are specific characteristics of jump-landings that are unique in dance (3).

Previous research has described dancers biomechanical landing pattern. An adequate jump-landing technique in dance is defined by starting and finishing in *demi-plié* position; meaning knee flexion, while heels contact the ground (Figure 3). The spring action of the *demi-plié* is essential in the jump-landing activities to prepare the jump and also to support the shock absorption mechanism during the landing phase (37). It also enables the use of the required *en dehors* while executing these movements (37).

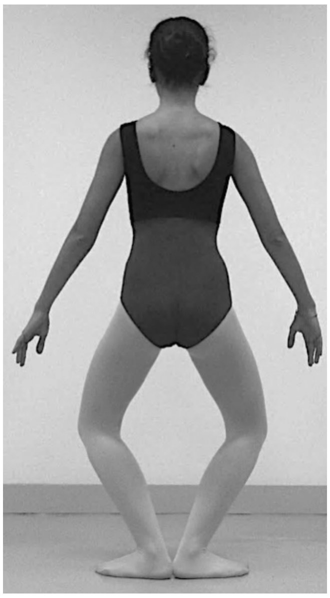

Figure 3. Depiction of a dancer in a *demi-plié* position (posterior view): ankle dorsiflexion, knee flexion, and external rotation of the lower extremities (*en dehors*), with the patella aligned with the second metatarsal.

From the *demi-plié* position, the jump begins with hip and knee extension, and foot-ankle plantarflexion (12, 37), transferring the weight from the heel to toes (12). The last point of contact with the ground, before the flight phase, are the toes to push-off. Maximum ankle plantarflexion (*pointé*) and alignment of the lower leg, ankle and foot are required during the flight phase (12, 66) preventing the sickling mechanism (presented in Chapter II – section 2.3.1). Still, in this phase, dancers have to maintain the lower extremities externally rotated (*en dehors*), which provides a medio-lateral work of the hip to assist with the alignment of the lower extremity during jump-landings (37). At the peak of the flight phase, the knees are extended and the ankles fully plantarflexed (12, 66). They also present an erect trunk posture during the entire jump-landing (13, 37). At the initial ground contact, the foot and ankle are plantarflexed; the first part of the body contacting the ground in the landing phase are the toes, and then the ball of the foot (12, 71). After the IC, the plantarflexed foot-ankle complex quickly and forcefully moves into a dorsiflexed position after the foot entirely contacts with the ground (71). Therefore, at IC, dancers land with nearly fully extended lower extremities; demonstrated by lower hip and knee flexion compared to other athletic populations, as well as maximum use of the ankle plantarflexion (13, 15). This pattern emphasizes the use of lower extremity joints excursion and also highlights the developed ability of the foot-ankle to roll throughout the foot (13, 57); this topic will be further addressed in Chapter II – section 2.3.1. Dancers then continue the landing phase by placing the heel on the ground coupled with knee flexion (*demi-plié*) to prepare for the next jump (12).

It is believed that the aforementioned extended landing pattern of dancers may hinder appropriate neuromuscular control to absorb the landing forces, which can potentially result in higher ground reaction forces (18, 22, 41, 68). This mechanism may increase the load on passive structures of the lower extremity joints (20), and consequently lead to a higher risk of injury. Yet, dancers appear to compensate the erect posture at initial contact with higher knee and ankle excursions during landing (13, 24). This biomechanical strategy allows the dissipation of the impact forces over a larger range of motion, and at the same time to achieve the desired aesthetics of a smooth and seemingly effortless landing in dancers (11, 13). This suggests that dancers appropriately take advantage of their extensive training of *demi-plié* movement. Also, dancers demonstrate decreased knee valgus compared to athletes (15, 24) that may contribute to a lower incidence of injuries (e.g., anterior cruciate ligament injury) in dancers (11). Despite the substantial research conducted in jump-landings, a comprehensive overview of lower extremity kinematics and kinetics in these tasks is still needed to comprehend the underlying patterns contributing to shock absorption mechanism in dancers and other athletic populations (known in Chapters III, IV, V, and VII as non-dancers).

2.2.2 Different types of jump-landings

It is noticeable that different tasks have been used in biomechanical studies investigating the lower extremity during jump-landing activities. A limitation of the biomechanical laboratory studies is that one cannot determine how the measured movement patterns relate to the biomechanics of real athletic motions (47). Nevertheless, an effort should be made to attempt to mimic athletic movements in the laboratory, as realistic as possible, increasing the ecological validity.

2.2.2.1 Drop landing and drop jump tasks

The drop landing has been primarily and extensively used to investigate lower extremity biomechanics (13-16, 19-22, 41, 46, 72). This task isolates the landing phase from the entire jump-landing sequence and may not take into account the preparation of a landing (45). The activation of the lower extremity musculature prior to landing affects joints’ kinematics and kinetics and potentially mediate loading (67). To drop to land (Figure 4) is reasonably different to performing a preceding jump before landing, as there is no dynamic motion prior to landing. Also, it is rarely performed outside the laboratory environments, and is considered as a less complex movement task (45). The landing phase of a drop landing and a spike jump have considerably differences in kinetic, kinematic and muscle recruitment strategies observed in volleyball players. Drop landings displayed a more extended landing pattern, most likely due to a more vertical body position (45). Thus, a drop landing does not mimic jump-landing tasks. Results from research using drop landing mechanics to model a landing phase of an entire jump-landing sequence should therefore be taken with caution (45, 47).

Additionally, previous research has primarily employed a drop vertical jump to assess injury risk (42, 68). This task is in a controlled environment, primarily vertical in nature, and may not fully represent the demands of various jumps (e.g., vertical vs. horizontal displacements) performed in many athletic and dance activities. For instance, there are substantial differences in the magnitude of knee kinematics and knee joint loadings between drop jumps and sidestep cutting tasks (47). In the sidestep cutting, athletes had lower knee flexion angle, higher knee valgus and internal rotation angles at IC, and higher knee moments (e.g., knee abduction moment was 6 times higher) (47). Sidestep cutting and drop jumps have different biomechanical motion patterns with the double-leg drop jump considered a more controlled and less complex task (47).

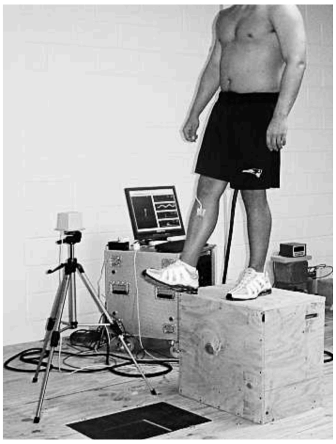

Figure 4. Demonstration of the starting position of a double-leg drop landing task (18).

2.2.2.2 Single- and double-leg landings

Double-leg landings can provide valuable information, however, may not represent a comprehensive view of an athletes' high-risk biomechanics (27); injuries often occur during single-leg landings (25, 47). This injurious mechanism is most likely due to a significantly more strenuous biomechanical demand on a single-leg compared with a double-leg (27). The impact forces, during single-leg landings, are solely absorbed by one lower extremity, with a decreased base of support and increased biomechanical demands from the trunk and pelvis (27). Moreover, single-leg landings have demonstrated lower sagittal hip and knee angles at IC and respective excursions compared to double-leg landing (20, 25).

Biomechanical magnitudes, especially joint moments, are greater during single-leg than double-leg landings, which suggests higher musculature demands needed to dissipate energy when landing in one limb (27, 44). Larger demands are required of posterolateral hip musculature in single-leg landings, to maintain a leveled pelvis and prevent excessive hip adduction and internal rotation (components of dynamic lower extremity valgus). Previous research have reported hip flexion and adduction moments in single-leg landings that were 2.2 and 6.6 times higher, respectively, than during double-leg landings (27). It is suggested that single-leg landings are challenging tasks that may enable better assessment of high-risk movement patterns (27). Also, different lower extremity energy dissipation strategies are adopted when performing a single- or double-leg landing, in sagittal and frontal planes (25). For example, in the sagittal plane, the hip and knee are the major contributors to energy dissipation in double-leg landings, whereas the hip and ankle primarily contribute during single-leg landings. Findings from previous research, exhibited that double-leg sagittal plane movements may not fully represent the neuromuscular strategies during high-risk activities (25, 27, 47). The biomechanics elicited during double-leg sagittal plane are not representative of biomechanics during more multidirectional sport-specific activities. Hence, single-leg and multidirectional tasks may complement scientific information (27).

2.2.2.3 Jump-landing directions

The different jump-landing directions meaningfully affect the lower extremity biomechanical strategies adopted to dissipate the impact forces during the landing phase (26, 27, 39, 40, 73). These biomechanical patterns suggest distinctive neuromuscular demands for each direction. For example, during the lateral jump-landing direction lower hip flexion, higher knee flexion, ankle dorsiflexion, and peak knee valgus angles were reported when compared to diagonal and forward directions (39, 40). Jump-landings in a frontal plane movement elicit more predominant signs of dynamic lower extremity valgus (27). Thus, using only the forward direction in jump-landing assessment may be misrepresenting essential aspects of neuromuscular control that are likely important to improve intervention programs of lower extremity injuries (73).

Multidirectional single-leg landings may better differentiate the biomechanical profiles in landings (26). Dancers also perform repetitive multidirectional landings, frequently supported in single-leg, with considerable horizontal displacement as part of their artistic and technical requirements. Thereby, this PhD research focused on multidirectional single-leg landings (Figures 5 and 6) of professional dancers and compared to non-dancers, to investigate the underlying biomechanical responses in different directions.

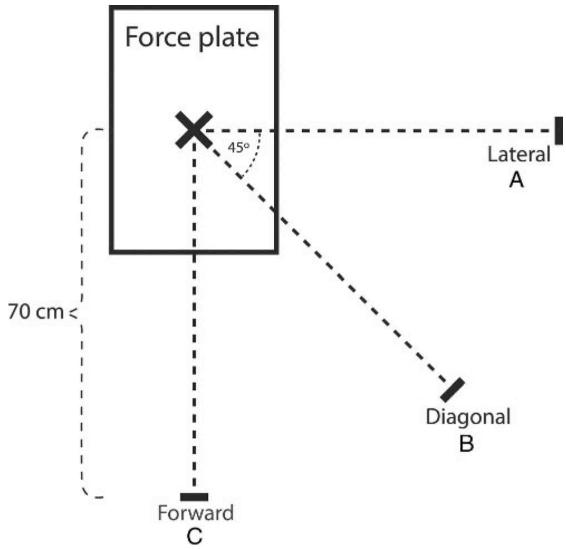

Figure 5. Representation of the three directions (A) lateral, (B) diagonal, (C) forward of the jump-landings performed in the studies of this PhD thesis.

Figure 6. Depiction of the single-leg jump-landing in the lateral direction, followed by a vertical jump-landing; marker set on the trunk and left dominant lower extremity. The highlighted image represents the landing phase analyzed in the studies of this PhD research.

2.3 Foot-ankle complex

The foot-ankle complex is a unique masterpiece of the human body. The complexity of the foot-ankle anatomy has a remarkable influence on its biomechanics (74). Unlike other segments of the lower extremity, the foot includes multiple bones and joints with a very intricate interaction (75, 76). This structure enables not only motion with a notable level of joint stability but is also a key-element in a wide variety of daily and functional activities (76). The movement of the foot-ankle complex joints is multiplanar and exceptional (37, 77, 78), which represents a challenge to define and characterize *in vivo* foot kinematics. Nonetheless, due to its important function, and also improvement of the accuracy of motion capture systems, the measurement of the foot kinematics has becoming more common (77).

The foot-ankle complex plays an important role to transfer forces and mechanical energy throughout the entire lower extremity, influencing its biomechanics due to the closed kinetic chain (19, 21, 28, 72). This structure provides a unique foundation to static and dynamic stability, mechanical leverage, shock absorption, and also to maintain balance (79). Additionally, it is remarkably energy efficient (79). Thus, all of the characteristics taken together demonstrates the natural resilience of the human foot (79). Thereby, the several joints of the foot and ankle have an outstanding position and function in the human body motion.

The interface between the lower extremity and the ground is the foot-ankle complex (74). The biomechanics of this structure is critical to the functionality of the lower extremity (28) in a wide variety of dynamic activities, including jump-landings. The individual joints and segments of the foot work simultaneously in a coordinated and synchronized system rather than isolated movements (76). The foot-ankle complex therefore can provide optimum forces attenuation and energy-absorbing characteristics (79) that must be investigated. Previous biomechanical studies in dancers jump-landings (13, 15, 16, 24) commonly modeled the foot as one rigid segment. This model neglects the intricate structure of the foot and ankle functional anatomy (77, 80, 81) and may attribute motion to one joint when in fact it occurs in more than one foot joints (82). Thus, biomechanical data concerning the foot-ankle complex kinematics using a multi-segmented foot model may provide further evidence related to dancers' foot-ankle structure.

2.3.1 Foot-ankle complex in dancers

Dancers often rely on the intricate foot-ankle complex while performing jump-landings. One of the main purposes of this structure is to initially attenuate the landing forces imposed on the body, while providing dynamic stability (19, 21, 76). A well-trained dancers' foot and ankle therefore are able to provide full support in landing tasks, and most likely work as an advantageous shock absorption mechanism (13, 19, 21, 37).

Dancers exhibit high motion in these structures due to the aesthetics and technique requirements. Professional dancers present 129% increased plantarflexion motion over the norm, which provides a total ankle range of motion 82% greater than norm (2). However, a standard goniometer was used to measure joint angles and ranges of motion (2). Additionally, dancers exhibited increased strength of the ankle plantarflexors; approximately 40% greater than the norm, which may be linked to the great and specific demands on the foot-ankle complex in dance (2). Although foot and ankle flexibility is of utmost importance to dancers aesthetics and technique, there is little evidence to support that extreme ankle range of motion is a predictor for subsequent injury (56).

There are some unique features to dancers, such as dancing *en pointé* (63). To dance *en pointé*, female dancers should have approximately 90º to 100º of plantarflexion in the foot-ankle complex joints (2). A considerable amount of plantarflexion is required in dancers, linked to strength and extensive mobility of different segments of the foot (1, 38, 83), and not just the ankle joint. The observed plantarflexion motion in dancers comes from the combined motion of different foot joints, such as talocrural, subtalar, and midtarsal joints of the foot-ankle complex (2, 83). Furthermore, it has been reported that during weightbearing *en pointé* and *demi-plié* positions in female *ballet* dancers, the talocrural joint is the greater contributor to plantar and dorsiflexion motions, with approximately 70% of the overall motion, measured using x-ray. Whereas the remaining 30% occurs in other foot joints (83). This suggests that when computing the foot-ankle complex using a single rigid segment to measure kinematics, it is likely providing insufficient information related to foot function as the other foot joints are also a contributor to the foot overall motion (77, 80, 81). This motion pattern in dancers is doubtless an effect of years of training technique (1, 2, 11, 13, 63) and most likely a result of some degree of skeletal molding in young dancers while bones are still maturing (2).

As previously described in Chapter II – section 2.2.1, dancers are frequently exposed to jump-landing activities with the foot-ankle complex having an important responsibility due to the toe-to-heel landing technique (13, 24). This structure function is vital to dancers' ability to perform, requiring flexibility and strength (1), which provides a distinctive landing pattern. The toe-to-heel technique, aforementioned in Chapter II – section 2.2.1, has been noticeably associated to adequately mitigate the landing forces (19, 41). Besides, the joint mechanics of the distal joints of the lower extremity are crucial to activate an adequate muscle pattern to contribute to shock absorption mechanism (67, 68, 84).

The talocrural joint anatomy is not a simple hinge joint (37, 85). In open kinetic chain, due to the oblique orientation of the talocrural axis, ankle dorsiflexion occurs with foot abduction and eversion; whereas plantarflexion occurs with adduction and inversion (37). This affects the position of the foot-ankle complex at IC. The described kinematically coupled foot-ankle inversion and adduction with plantarflexion is known as the sickling mechanism. Dancers are subject of extensive foot-ankle training, to work on the required alignment of the lower leg, ankle and foot, through the evertors and plantarflexors muscles (37, 66). This helps avoiding the sickling mechanism in landings. Simultaneously, they have to externally rotate the lower extremities, through the activation of the hip external rotators muscles (37). Thereby, the described strategy can be valuable for dancers since it may assist to prevent some injuries to the ankle (e.g., ankle or midfoot sprains). This suggests that learned coordination patterns at the foot-ankle joints may also be important to prevent injuries (66). Consequently, due to the paucity of research examining dancers multi-segmented foot in jump-landing activities, there is a gap in dance science related to such an important, intricate and often requested structure in dance.

2.3.2 Multi- vs single-segmented foot models

Recent research has shown that there is clear evidence of how selection of marker data from different bones can influence the kinematic description of a segment model (82). Several biomechanical studies investigating lower extremity in jump-landing activities, have modeled the foot as a single rigid segment (13, 15, 19, 20, 24, 41, 51, 53). The bones within each segment of the foot move relative to each other (82). Thus, this model provides limited kinematic data since it neglects information about the relative motion between and within the different foot segments (77, 78, 80, 81). The kinematic outputs of the talocrural joint, when using the single-segmented model, also includes motion of other foot joints (e.g., forefoot-hindfoot). The single-segmented foot model simplifies the intricate structure of the foot-ankle complex and does not provide meaningful information related to the foot kinematics (80, 81). Previous research has reported distinctive ankle kinematics during walking and landing activities using different foot-ankle model approaches (multi- vs single-segmented models), with higher ankle magnitudes in the single-segmented foot model (80, 81).

The different joints of the foot produce a challenging process to define and represent *in vivo* foot kinematics using 3D motion capture system. Recently, different models to compute the foot into multiple segments have been developed and proposed. Multi-segmented foot models divide the foot into different segments to obtain meaningful data about its function (75). The Oxford Foot Model is one of the multi-segmented foot models that have been employed to investigate the foot-ankle complex (77, 78, 81, 86-88). It presents kinematic information of the hindfoot-tibia, forefoot-hindfoot and hallux-forefoot joints of the foot-ankle complex (77, 78, 86). Thereby, representing a three-segment foot (hindfoot, forefoot and hallux), and tibia (Figures 7 and 8) (77, 78, 86).

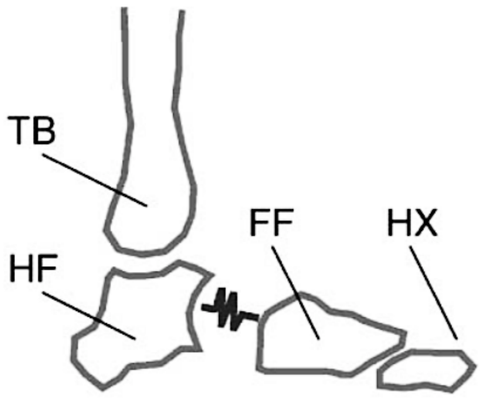

Figure 7. Representation of the three-segment foot of the Oxford Foot Model. Segments: TB = Tibia; HF = Hindfoot; FF = Forefoot; HX = Hallux (86).

This model has been used over years, and its reliability has been reported in adults and children, healthy and pathological populations, performing different tasks, such as walking, running or step down (77, 78, 81, 87, 88).

Figure 8. Anterior view of the foot and ankle markers placement according to the Oxford Foot Model.

Remarkably, the foot-ankle complex is paramount to dancer’s technique and aesthetics; playing an important role to adequate mitigate the landing forces (19, 21, 41). Foot modeling as a multi-segment can provide an insight of the foot kinematics during jump-landings to better understand the intricate foot-ankle complex abilities. Due to the considerable range of motion in the foot-ankle complex, its functional anatomy, and required use in dance, research modeling the foot as a multi-segment during dynamic tasks is relevant and needed in dance.

2.4 Computational modeling and simulation

OpenSim software was chosen to further explore and employ computational modeling on the experimental data collected for this PhD research. It is an open-source platform for modeling, simulation and analysis of the neuromusculoskeletal system (32, 33). OpenSim generates computational simulation of human movement allowing two primary tasks (31, 32):

1) to calculate variables that are difficult to measure experimentally, such as muscle forces and joint loads for a wide range of scenarios – activities of daily living and athletic performances;
2) to predict and examine movement from models of motor control, such as kinematic adaptations of human gait.

Computational modeling and simulation also permit to analyze musculoskeletal systems through the study of neuromuscular coordination and joint loads estimation, which helps researchers to gain valuable insights into human movement (31, 33). Additionally, OpenSim uses the global optimization method, which is highly dependent on joint kinematic constraints to remove the possibility of joint non-anatomical positions (89). It has been considered has an efficient and reliable method for calculation of the position and orientation of the model (89). This method has the potential to minimize soft tissue artifact (90), a known concern in biomechanics research. Even when retro-reflective markers are carefully placed on selected anatomical landmarks of the participant, soft tissue artifact can affect the outputs, leading to misinterpretations (90).

2.4.1 OpenSim: a computational modeling and simulation approach

OpenSim has a wide range of capabilities for generating and analyzing musculoskeletal models and dynamic simulations. It also provides flexibility and control to users when inputting data (e.g., experimental measurements that drive a simulation) and outputting results of interest (e.g., joint angles and muscle forces) (32). The first step is to determine which model to use to guide the analysis of kinematics and dynamics (31). Thus, the first step is the development of a biomechanical model. This model includes a set of mathematical equations that describes a physical system representing human neuromuscular system that acts on a rigid multibody skeletal structure (31, 32). A model also represents the dynamics of rigid bodies and joints that are acted upon forces to produce motion. It is defined by the components of the model (e.g., rigid bodies and musculotendons), their properties (e.g., body masses and muscle fiber lengths), and the connections among components (e.g., joints) (32).

The elements of the musculoskeletal system are modeled by sets of differential equations that describe muscle contraction dynamics, musculoskeletal geometry and body segmental dynamics (31, 33). The musculoskeletal model is then scaled to match the participant anthropometry. The dimensions of each body segment in the model are scaled based on relative distances between pairs of markers obtained from motion capture system and the corresponding virtual marker locations in the model. In addition, the mass properties of the body segments are scaled proportionally so that the total measured mass of the participant is reproduced, as well as the muscle fiber lengths and tendon slack lengths of the muscle-tendon actuators (33). Subsequently, inverse kinematics is solved to estimate the model joint angles and translations to best reproduce the experimental marker trajectory. This step is formulated as a least-squares problem that minimizes the differences between the measured marker locations and the model's virtual marker locations (33).

It is recommended to perform inverse kinematics and dynamic analyses using a model that realistically represents physiological joints and is scaled to the anthropometry of the participant. The next two steps include: 1) residual reduction algorithm, which is applied to make the model generalized coordinates more dynamically consistent with the experimental ground reaction forces and moments; and 2) static optimization or computed muscle control, to generate a set of muscle excitations that produce a coordinated muscle-drive simulation of the participant's movement (33). At the end of this process, useful information of lower extremity muscle forces and activation profiles are provided. The use of this approach in jump-landings is extremely helpful to identify the causes of neuromusculoskeletal impairments and how it affects the movement itself.

2.4.2 Challenges presented in computational modeling and simulation

The development of computational modeling and simulation of the human musculoskeletal movement is not only a great challenge, but also a complex and multistep process (31, 33). The use of OpenSim to study daily activities and athletic performance has increased in the biomechanics field over the past decades (31, 32). Nevertheless, it has been observed that lack of best practices for the verification and validation of musculoskeletal models and simulations, which is a challenging process, remains a major barrier to wider use and impact (31).

A dynamic model of the musculoskeletal system and its interaction with the environment is needed to create movement simulations (33). However, based on OpenSim guidelines there is no scientific consensus on when a simulation is "good enough". OpenSim inverse kinematics is an optimization method in movement analysis which fits a multi-segment biomechanical model to a set of marker positions measured experimentally (35). The accuracy of inverse kinematics depends on the exact placement of the markers and on the maximum markers error (35). The main objective is that the measured marker locations fit those of the model's virtual markers (33) (Figure 9). The recommended fitting error upper bound of OpenSim inverse kinematics is to have the maximum marker error and the maximum Root Mean Square (RMS) error below 0.04 and 0.02m, respectively. This is accomplished with marker set optimization; which is a subjective process due to the manual markers' adjustments and depends on users' skills and experience. It is common to alternate between the scale and inverse kinematics tools to fine-tune segment dimensions and marker positions that yield low marker errors for the performed task. Consequently, developing a subject-specific musculoskeletal model and simulation of movement has represented a time-consuming process (91).

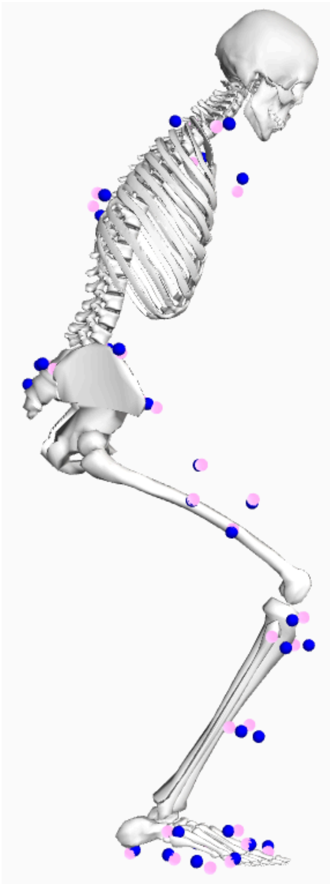

Figure 9. Illustration of the sagittal view experimental (blue) and virtual (pink) markers, after running OpenSim inverse kinematics.

During all of these processes, visualization of models and simulations are critical for interpreting, troubleshooting and communicating results (32). It also becomes noticeably that the sensitivity of musculoskeletal models to the modeler decision should be a concern in computational modeling (30). On one hand, it enables experts to improve models, while on the other hand, less experienced modelers may make incorrect conclusions (34). Furthermore, outputs from musculoskeletal simulations are affected by measurement error and model parameter uncertainties, which are important to consider when interpreting results (36). Variations in model predictions (e.g., joint angles, joint moments and muscle forces) due to uncertainties (operator-dependent or related to unavailability of measurements *in vivo*) and errors in marker positions are a trait of musculoskeletal modeling approaches (34, 91). In this process, it is the user's responsibility to ensure that the models and simulations realistically represent the experimental data and findings from the simulations are trustworthy (30, 31).

CHAPTER III

3. Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings

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Abstract

Purpose: To compare lower extremity biomechanics between professional dancers (PDs) and non-dancers (NDs) during multidirectional single-leg landings. **Methods:** Fifteen PDs (27±7 years, 1.69±0.1m, 57.8±9.3kg), and 15 NDs (25±5 years, 1.69±0.1m, 66±10.2kg) performed single-leg jumps in three directions: 1) lateral, 2) diagonal, and 3) forward. Dominant LE biomechanical data were collected using a motion capture system. Data were processed in Visual3D. LE kinematic (hip, knee and ankle joint angles in sagittal and frontal planes, and range of motion [ROM]) and kinetics (hip and knee internal joint moments and vertical ground reaction force) variables were analyzed at initial contact (IC), peak vertical ground reaction force (PvGRF), and peak knee flexion (PKF). Repeated measures ANOVAs were conducted ($p<0.05$). **Results:** At IC, statistically significant interactions were found for ankle frontal and hip sagittal angles ($p<0.05$). Main effects for groups and jump directions were attained ($p<0.05$). PDs at IC had lower hip and knee flexion, and higher ankle plantarflexion than NDs. PDs had significantly higher knee (PDs: 41±6.1; NDs: 33.8±8.4) and ankle (PDs: 53.7±3.4; NDs: 38.9±8.9) ROM than NDs. At IC, the lateral jump had a higher hip abduction moment, hip abduction and ankle inversion, and lower hip flexion and ankle plantarflexion than the forward and diagonal jumps. The LJ (15.5±7.7) had higher hip excursion than the FJ (12.7±5.4). **Conclusion:** PDs higher extended posture at initial contact promoted an efficient use of the knee and ankle ROM to dissipate the landing forces. Regardless the group, jump directions also solicited different biomechanical responses, particularly between lateral and forward directions. These strategies should be considered for implementation in prevention programs, as it can foster adequate lower extremity neuromuscular control.

Keywords: Lower extremity; Kinematics; Kinetics; Dancers; Multidirectional landings.

Introduction

Lower extremity (LE) musculoskeletal injuries have a multifactorial etiology (1). More than 60% of all dancers’ injuries occur to the LE (2), similar to other athletic populations (3, 4). However, 66% to 72% of the injuries in dance are overuse in nature (2, 5), mainly because dancers do not stop dance activities, thus less serious injuries may progress to overuse injuries. Consequently, only 28% to 34% of dance injuries are classified as acute (2, 5), whereas acute injuries have been reported at 70% in other athletic populations

(6). The common type of injuries in dance are tendinosis (32%), ligament sprain (17%), and muscle strain (15%), commonly linked with stress related factors; also, those are the three conditions that required greatest amount of treatment (2). The injuries have short and long-term health consequences that affect performance and quality of life with potential for extended disability (7). Intrinsic and extrinsic risk factors have been proposed (1, 8). Deficits in muscle strength and/or muscle imbalance (9), decrease in neuromuscular control (9-11), and poor technique (8) are risk factors frequently associated to LE injuries. Injury prevention programs have been shown to be effective in altering modifiable risk factors and decreasing injury rates in the specific study samples (12). Still, overall LE injury rates have remained steady across multiple populations (3), suggesting that injury prevention programs need broader implementation. Identifying and understanding the modifiable musculoskeletal injury risk factors can lead to improved primary prevention programs (7, 8).

Substantial research has been done to investigate modifiable risk factors during jump-landing activities (10, 11, 13-17). LE role during landing is to reduce and control the downward momentum acquired during the flight phase, which places a strenuous demand on the LE (18). It has been reported that a more flexed landing pattern, with higher trunk, hip and knee flexion during the landing phase is associated with higher shock absorption and consequently lower ground reaction force (11, 19). Contrastingly, it has been reported that lower flexion angle during landing augments the reliance on the frontal plane motion and loads to decelerate the body center of mass (20). Though, excessive frontal plane motion and loading is a common knee injury mechanism (10).

Previous research reported that dancers have a more erect trunk and LE biomechanical pattern during the landing phase compared to non-dancers and other athletes (13, 21). This extended pattern may hinder appropriate neuromuscular control to absorb the landing forces, resulting in higher ground reaction forces (11, 14, 17, 19, 22), potentially increasing the risk of injuries. Still, dancers compensate the extended position at initial contact with higher knee and ankle excursion (13) as a strategy to dissipate landing forces over a higher range of motion (17). Also, they may employ a muscular strategy, through higher plantarflexor muscle action, to attenuate the landing forces (11, 14, 17, 22) without a concomitant increase of PvGRF (17). Further, dancers perform repetitive single-leg multidirectional landings with significant horizontal displacement as part of their artistic requirements. The drop landing task has been primarily used to investigate lower extremity biomechanics. Yet, some limitations have been suggested, as it inhibits the isolation of the landing phase from the entire jump-landing task limiting the preparation for landing, and it has been implied as a less complex movement (23). Additionally, literature reports that LE kinematic and kinetic parameters are significantly affected by the jump-landing direction which is absent in a drop landing task (16).

During lateral jump-landing lower hip flexion, higher knee flexion, ankle dorsiflexion, and peak knee valgus angle were reported when compared with other directions (16, 24). The different landing directions require distinct biomechanical strategies to dissipate the impact forces (15, 25, 26). Furthermore, those impact forces, during single-leg landings, are solely absorbed by one LE, with a decreased base of support and increased biomechanical requirements from the trunk and pelvis (15), with lower knee and hip range of motion (27). Thereby, the body needs to attenuate the impact distally (27), suggesting that further investigation of the distal and proximal joint control during landing is warranted. It is relevant to consider and incorporate jump-landing direction skills during intervention programs. Also, understanding the biomechanical landing patterns in professional dancers can contribute to comprehend what may facilitate shock absorption during landing. To our knowledge, there is limited research investigating the biomechanical responses during multidirectional single-leg landings between professional dancers and non-dancers that can contribute to the differences in acute injury rates (28). Therefore, the purpose of the current study was to investigate kinematic and kinetic differences between professional dancers and non-dancers, during multidirectional (lateral, diagonal, forward) single-leg landings. It was hypothesized that 1) professional dancers would have a more extended landing posture (lower hip and knee flexion, and higher plantarflexion), higher ankle excursion and peak vertical ground reaction force during single-leg landing compared to non-dancers, and 2) lateral jump-landing would have higher knee and lower hip flexion compared with the forward and diagonal jump directions.

Methods

Participants

A single-session descriptive group comparison design was used. An a priori sample size estimation, related to landing techniques between dancers and athletes and lower extremity biomechanics (29, 30) with an effect size of 0.8, an exploratory alpha level of 0.05, revealed that a minimum of twenty-four subjects would be required to achieve 80% statistical power. A total of 30 participants, 15 professional dancers (26.6 ± 7 years, 1.69

± 0.1 m, 57.8 ± 9.3 kg), and 15 non-dancers (25.0 ± 5 years, 1.69 ± 0.1 m, 66.0 ± 10.2 kg), volunteered to participate in the current study. All data were collected on the dominant side of each participant. The dominant side was defined as the preferred single-leg landing after performing a countermovement jump (31). To be included in this study, all participants were required to be between 18 and 40 years old, and physically active; furthermore, professional dancers (ballet and modern dance) had to practice a minimum of 10h/week of dancing, whereas non-dancers had to exercise a minimum of 3h/week and had different sports background (basketball, running, surf, gymnasium, swimming, football, tennis, boxing). Participants were excluded if they had a recent history of lower extremity injuries, a history of lower extremity surgery within the past five years, any recent pain to the lower extremity that would impair the ability to jump or any known neurological or cognitive disorder. All procedures performed in the study were in accordance with the ethical standards of the Institutional Ethical Review Committee and with the 1964 Helsinki declaration and its later amendments. An informed consent was obtained from all participants prior to the study.

Instrumentation & Experimental Procedures

Lower extremity biomechanical data were collected using a 10-camera high-speed three-dimensional motion capture system (Opus, Qualisys AB, Gothenburg, Sweden) sampling at 200 Hz. A Bertec force plate (FP4060-10, Bertec Corporation, Columbus, Ohio) recorded ground reaction force data, sampling at 1000 Hz.

Participants' demographics and anthropometrics measures were obtained by the same researcher (AMA). Thirty-three retro-reflective markers were placed on selected body landmarks, and a four markers cluster was used on the thigh of the dominant side. The retro-reflective markers were secured on body landmarks using double-sided tape, and the marker cluster was placed with a velcro band around the participant's thigh. Participants stood barefoot and wore spandex shorts. Women also wore a sports bra during testing. A 5-min self-directed warm-up period was provided. Subsequent to the warm-up, a static trial was collected. The four calibration markers (medial knee and malleolus, upper posterior calcaneus, and first distal metatarsal) were removed prior to the recording of the jump-landing trials.

The same researcher (AMA) provided instructions to each participant on how to perform each jump. Participants stood on the nondominant leg, 70 cm away from the center of the force plate then proceeded to randomly perform multidirectional single-leg jump-landings: (A) lateral (LJ), (B) diagonal (DJ), and (C) forward (FJ) (Figure 10). Upon landing, participants immediately transitioned into a maximal vertical jump, which was followed by a second landing also in the middle of the force plate, on the dominant leg (Figure 11) (15, 16, 24, 25). They were asked to perform three trials for each direction. Between each trial, 30-seconds rest was provided. A trial was deemed successful when participants jumped into the center of force plate, kept their hands on the hips, and did not: (i) lose balance, (ii) touch the force plate with the non-dominant foot, and/or (iii) did not hop or adjusted the landing foot upon contact with the force plate. Unsuccessful trials were defined as the loss of balance, hopping or stepping off the force plate or using their arms to maintain balance. For the present study, only the first landing was analyzed.

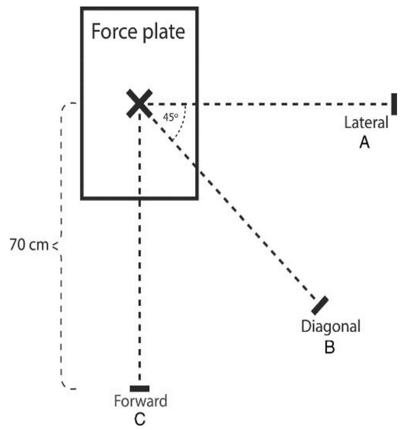

Figure 10. Representation of the jump-landing directions: (A) lateral (LJ); (B) diagonal (DJ); (C) forward (FJ). Distance between starting point and center of force place for each jump was 70cm.

Figure 11. Visual illustration of the task (e.g., lateral direction): single-leg jump-landing, followed by vertical jump.

Data Processing

Markers were manually identified using QTM (Qualisys Track Manager) software, and then were exported to and processed in Visual 3D (C-Motion, Inc, Rockville, USA). A laboratory coordinate system was established with the positive x-axis, y-axis, and z-axis represented forward, left, and upward directions, respectively. From the static trial, a kinematic model was used to quantify joint motion, composed of four reconstructed segments; foot, shank, thigh of dominant extremity, and the pelvis for each participant. Knee and ankle joint centers were determined as the midpoints between the medial and lateral epicondyles and malleoli markers, respectively. The hip joint centers were estimated using a previously reported regression equation (32).

Force plate data were used to defined the IC point, at which the vertical impact force plate surpassed a threshold of 10N (15, 26, 33). The landing phase was defined as the time interval between the IC with the force plate and peak knee flexion (PKF) (13, 21, 33). Peak vertical ground reaction force (PvGRF) data was defined as the vertical ground reaction force maximum value during the landing phase. The kinematic variables of interest were: sagittal and frontal planes joint angles for hip, knee, and ankle. Hip, knee and foot joint excursion in sagittal plane was computed as the subtraction of the angle between IC and PKF during the landing phase (13). All joint angles were reported in degrees (º). The kinetic variables calculated were: internal joint moments for hip and knee, and vertical ground reaction force. Internal joint moments were computed using conventional inverse dynamics analysis (34) and normalized to each participant's body mass and height (Nm/kgm) (26), representing the internal load applied to each joint. The ground reaction force was normalized to body weight (BW). Residual analysis on joint kinematics and kinetics was conducted to determine the optimum cut-off frequency (34). A 4th order low-pass Butterworth filter with a 10Hz cut-off was employed on joint kinematics and kinetics. All dependent variables were measured at IC, PKF, and PvGRF. Additionally, time to each peak was also computed.

Statistical Analyses

All statistical analyses were performed using SPSS (IBM, Chicago, IL). Descriptive statistics and normalcy tests were conducted. A 2 (group) x 3 (jumps) repeated-measures analysis of variance (ANOVA) was conducted to determine interactions between groups and jumps, as well as main effects between the jumps (LJ, DJ, and FJ) and groups (professional dancers and non-dancers) for all dependent variables. Pairwise comparisons with a Bonferroni adjustment were performed when significant differences were observed. The alpha level for statistical significance was set at $p<0.05$ for all data.

Results

Descriptive statistics (means ± standard deviations) of the kinematic and kinetic dependent variables, for the two groups and the three jump directions are summarized in table 1. An interaction effect between groups and jumps was significant for ankle, and hip angles in frontal and sagittal planes, respectively, at IC ($p<0.05$). Professional dancers during the forward jump-landing showed higher ankle eversion than non-dancers during diagonal and lateral directions. Moreover, professional dancers performing the lateral jump-landing had lower hip flexion than during their forward direction, and the three landing directions of non-dancers. Further, professional dancers during the diagonal jump-landing also presented lower hip flexion compared to non-dancers' diagonal and forward directions. A statistically significant main effect was found for group ($p=0.008$), as well as, for jump directions ($p<0.05$).

For group main effect, professional dancers at initial contact had lower hip and knee flexion, and higher ankle plantarflexion than non-dancers (Figure 12).

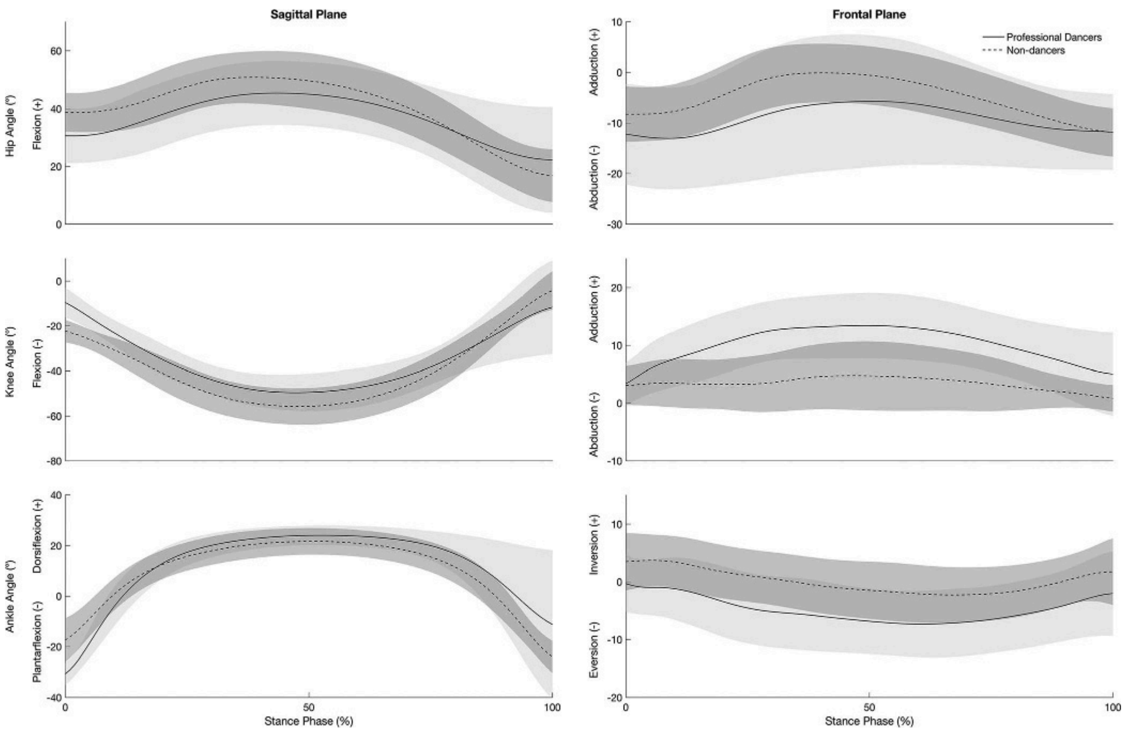

Figure 12. Comparisons of mean (SD) of hip, knee and ankle sagittal and frontal plane angles observed between professional dancers and non-dancers. Left column illustrates aggregated sagittal plane data, and the right column depicts ensemble frontal plane data.

Table 1. Descriptive table (mean ± standard deviation) of the kinematic (angles in degrees) and kinetics (joint moments in Nm/Kgm) variables at IC, PvGRF and PKF.

	Initial Contact				Peak Vertical Ground Reaction Force				Peak Knee Flexion			
	Lateral Jump	Diagonal Jump	Forward Jump	Total	Lateral Jump	Diagonal Jump	Forward Jump	Total	Lateral Jump	Diagonal Jump	Forward Jump	Total
Joint Angles:												
Hip Flexion (+)												
Professional Dancers	25.7±4.3	31.9±5.7	35.2±5.7	30.9±5.2*	38.6±5.7	41.2±5.1	43.7±5.9	41.2±5.6	43.6±6.4	47.5±7.0	48.9±6.4	46.7±6.6
Non-Dancers	35.6±7.4	39.5±6.2	40.9±4.7	38.7±6.1*	43.7±8.4	44.5±7.6	46.2±8.3	44.8±8.1	48.9±9.8	50.0±8.0	51.8±9.5	50.2±9.1
Total	30.7±7.8*	35.7±6.0*	38.0±5.9*		41.2±7.5	42.8±6.6	45.0±7.2[c]		46.3±8.6[a]	48.7±7.5	50.3±8.1	
Knee Flexion (-)												
Professional Dancers	-13.2±5.4	-14.1±5.9	-12.4±5.1	-13.2±5.5*	-46.8±5.6	-43.1±5.7	-45.0±7.3	-45.0±6.2	-53.7±5.8	-55.0±5.6	-54.3±4.4	-54.3±5.2
Non-Dancers	-24.2±4.8	-22.4±3.9	-20.3±3.9	-22.3±4.2*	-47.9±8.6	-43.7±9.7	-42.0±10.3	-44.5±9.5	-56.4±8.7	-56.3±6.0	-55.6±8.3	-56.1±7.7
Total	-18.7±7.5	-18.2±6.5	-16.3±6.0[c]		-47.3±7.1[a]	-43.4±7.8	-43.5±8.9		-55.0±7.4	-55.6±5.7	-55.0±6.6	
Ankle Plantarflexion (-)												
Professional Dancers	-28.1±4.5	-30.3±4.2	-31.3±4.1	-29.9±4.2*	23.8±3.0	17.5±3.4	17.2±6.0	19.5±4.1*	26.4±3.1	23.3±3.0	21.6±3.2	23.8±3.1
Non-Dancers	-14.1±8.5	-18.7±7.3	-19.3±8.4	-17.4±8.0*	21.6±4.5	15.3±5.0	11.9±5.0	16.3±4.9*	24.8±5.0	21.4±3.6	18.5±4.4	21.6±4.3
Total	-21.1±9.7[a]	-24.5±8.3	-25.3±8.9		22.7±3.9[a]	16.4±4.4	14.6±6.1		25.6±4.2*	22.3±3.4*	20.0±4.1*	
Hip Abduction (-)												
Professional Dancers	-9.9±3.3	-6.3±4.4	-1.5±5.4	-5.9±4.4	-2.3±5.1	-1.1±5.0	2.4±4.9	-0.3±5.0*	1.0±5.8	2.2±5.3	5.6±5.6	3.0±5.6
Non-Dancers	-11.8±3.5	-9.1±3.7	-3.5±4.2	-8.2±3.8	-5.2±3.7	-4.8±4.7	-0.6±4.6	-3.5±4.3*	-2.3±5.0	-1.2±4.8	2.6±4.3	-0.3±4.7
Total	-10.9±3.5*	-7.7±4.2*	-2.5±4.8*		-3.7±4.6	-3.0±5.2	0.9±4.9[c]		-0.6±5.6	0.5±5.3	4.1±5.1[c]	
Knee Adduction (+)												
Professional Dancers	2.3±3.2	3.0±3.7	3.9±4.0	3.1±3.7	10.8±5.0	10.5±5.3	12.0±4.5	11.1±4.9*	12.1±6.1	13.3±5.6	14.2±4.6	13.2±5.5*
Non-Dancers	2.9±3.5	2.9±3.0	3.3±3.3	3.0±3.3	3.4±4.3	4.2±4.8	4.5±4.3	4.0±4.4*	4.5±5.6	5.0±5.8	5.2±5.7	4.9±7.6*
Total	2.6±3.3[b]	2.9±3.3	3.6±3.6		7.1±5.9	7.4±5.9	8.2±5.8[c]		8.3±6.9[b]	9.1±7.0	9.7±6.9	
Ankle Inversion (+)												
Professional Dancers	2.7±4.8	0.4±4.3	-2.3±4.4	0.3±4.5	-4.7±4.1	-4.0±5.2	-7.5±4.2	-5.4±4.5*	-5.6±3.5	-5.5±5.0	-7.8±4.6	-6.3±4.3*
Non-Dancers	4.8±4.7	3.6±5.2	2.1±3.8	3.5±4.6	1.9±5.1	0.8±3.8	-2.2±2.8	0.2±3.9*	0.6±4.0	-0.9±3.3	-3.5±4.7	-1.3±4.0*
Total	3.8±4.8*	2.0±5.0*	-0.1±4.6*		-1.4±5.7	-1.6±5.1	-4.8±4.4[c]		-2.5±4.9	-3.2±4.7	-5.7±5.0[c]	
Joint Moments:												
Hip Extension (-)												
Professional Dancers	-0.071±0.053	-0.044±0.072	-0.039±0.077	-0.051±0.067	-0.69±0.23	-0.57±0.22	-0.52±0.23	-0.59±0.23	-0.77±0.28	-0.73±0.31	-0.58±0.23	-0.7±0.3
Non-Dancers	-0.024±0.086	-0.040±0.083	-0.061±0.090	-0.042±0.086	-0.56±0.39	-0.44±0.34	-0.38±0.34	-0.46±0.36	-0.79±0.38	-0.68±0.40	-0.64±0.45	-0.7±0.4
Total	-0.048±0.075	-0.042±0.076	-0.050±0.083		-0.62±0.32[a]	-0.50±0.29	-0.45±0.30		-0.78±0.33[b]	-0.71±0.36	-0.61±0.35	
Knee Flexion (-)												
Professional Dancers	-0.065±0.046	-0.05±0.040	-0.045±0.041	-0.053±0.042	1.24±0.21	1.17±0.38	1.30±0.43	1.31±0.3	1.43±0.23	1.59±0.35	1.60±0.31	1.54±0.3
Non-Dancers	-0.050±0.065	-0.046±0.065	-0.046±0.070	-0.047±0.067	1.36±0.48	1.25±0.53	1.32±0.49	1.24±0.5	1.56±0.34	1.76±0.38	1.79±0.35	1.70±0.4
Total	-0.058±0.056	-0.048±0.053	-0.046±0.056		1.30±0.37	1.21±0.45	1.31±0.45		1.50±0.29[a]	1.68±0.37	1.69±0.34	
Hip Abduction (-)												
Professional Dancers	-0.044±0.100	-0.014±0.076	0.031±0.051	-0.009±0.076	-1.09±0.24	-1.00±0.33	-0.93±0.25	-1.01±0.27*	-1.23±0.33	-1.21±0.38	-1.11±0.32	-1.20±0.34*
Non-Dancers	-0.014±0.054	0.014±0.054	0.039±0.066	0.013±0.058	-0.85±0.29	-0.73±0.28	-0.70±0.30	-0.76±0.29*	-0.95±0.22	-0.91±0.20	-0.90±0.22	-0.90±0.21*
Total	-0.029±0.083*	-0.0001±0.66*	0.035±0.058*		-0.97±0.29[b]	-0.86±0.34	-0.82±0.30		-1.09±0.31[b]	-1.06±0.33	-1.00±0.29	
Knee Adduction (+)												
Professional Dancers	-0.0014±0.040	0.012±0.033	0.031±0.026	0.014±0.033	-0.73±0.26	-0.71±0.34	-0.63±0.26	-0.69±0.29*	-0.73±0.31	-0.81±0.35	-0.73±0.31	-0.76±0.32*
Non-Dancers	0.0010±0.024	0.0034±0.021	0.014±0.023	0.006±0.023	-0.45±0.20	-0.42±0.22	-0.33±0.22	-0.40±0.22*	-0.48±0.23	-0.48±0.24	-0.44±0.28	-0.47±0.25*
Total	-0.0002±0.032	0.008±0.028	0.022±0.026[c]		-0.59±0.27	-0.57±0.32	-0.48±0.28[c]		-0.61±0.30	-0.64±0.34	-0.59±0.33	

*Statistically significant between three jump-landing directions or between the two groups ($p<0.05$);
[a] Statistically significant compared with diagonal and forward ($p<0.05$);
[b] Statistically significant compared with forward ($p<0.05$);
[c] Statistically significant compared with diagonal and lateral ($p<0.05$).

Professional dancers, at PvGRF, also demonstrated lower hip abduction angle, and higher hip abduction moment, knee abduction moment, and ankle dorsiflexion and eversion angles than non-dancers. A higher hip and knee abduction moment, knee adduction, and ankle eversion in professional dancers was also observed at PKF. Further, professional dancers had significantly higher knee (PDs: 41±6.1; NDs: 33.8±8.4) and ankle (PDs: 53.7±3.4; NDs: 38.9±8.9) excursion than non-dancers. Professional dancers took longer (0.21±0.04) than non-dancers (0.18±0.04) to achieve peak knee flexion.

There was a significant main effect for jumps (Figure 13). In particular, at initial contact, the LJ had a higher hip abduction moment, hip abduction and ankle inversion, and lower hip flexion and ankle plantarflexion than the forward and diagonal jumps, and lower knee adduction angle than the forward jump. Further, the DJ exhibited a higher hip abduction, hip abduction moment, and ankle inversion, and lower hip flexion than the FJ. It was also observed that the FJ presented higher knee adduction moment and lower knee flexion than the other two jumps, and a higher knee adduction than the LJ.

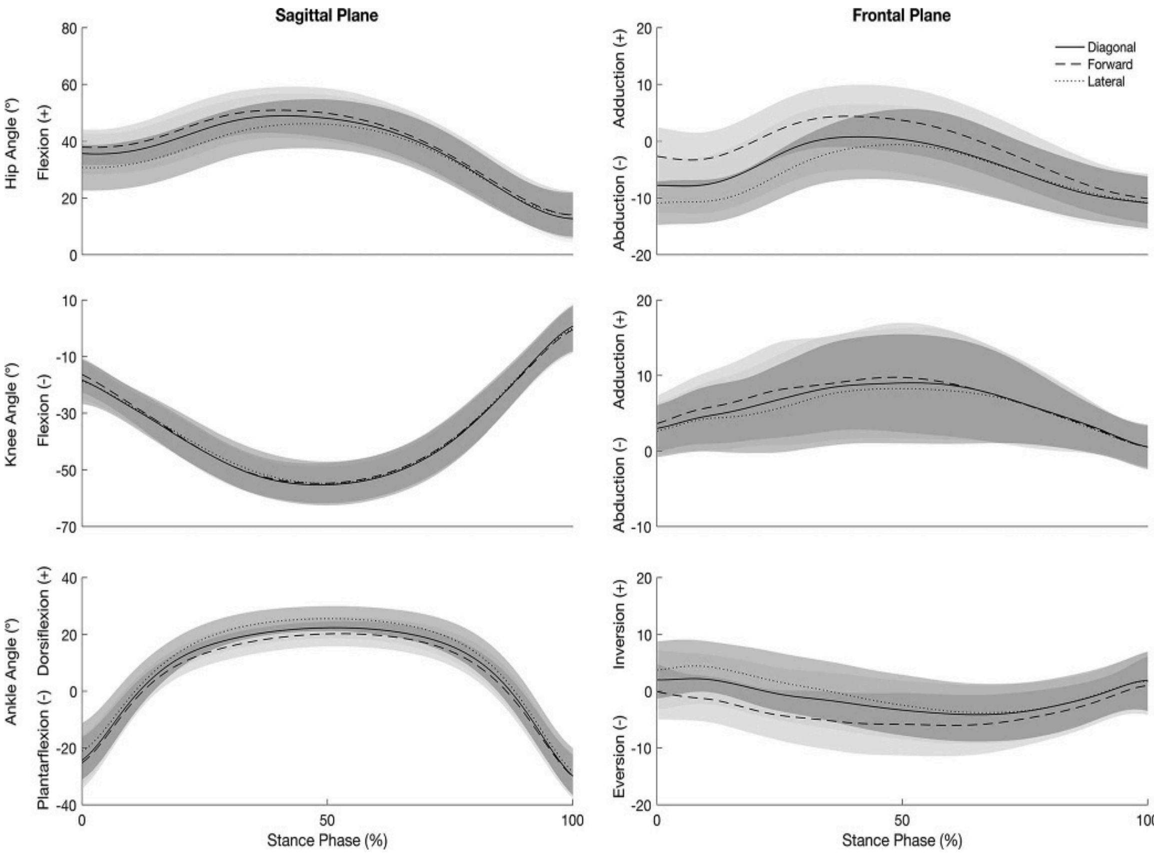

Figure 13. Comparisons of mean (SD) of hip, knee and ankle sagittal and frontal plane angles observed among multidirectional jumps. Left column illustrates aggregated sagittal plane data, whereas the right column depicts ensemble frontal plane data.

At PvGRF, the FJ demonstrated higher hip flexion, hip adduction, knee adduction and ankle eversion, and lower knee abduction moment when compared to the other two jumps. The LJ was significantly higher for hip extension moment, knee flexion, and dorsiflexion than the other two jumps, and had a higher hip abduction moment than the FJ.

At peak knee flexion, the LJ demonstrated lower hip flexion and knee extension moment and higher ankle dorsiflexion compared to DJ and FJ, and the DJ ankle dorsiflexion was also higher than the forward jump. The FJ exhibited a higher hip adduction angle compared to the other two, and higher knee adduction angle and lower hip extension and abduction moment when compared to LJ.

Lastly, the LJ took longer to peak knee flexion than FJ (LJ: 0.21±0.05; FJ: 0.19±0.04), and to PvGRF than the two other jumps (LJ: 0.18±0.09; DJ: 0.13±0.06; FJ: 0.14±0.07). The LJ (15.5±7.7) had higher hip excursion than the FJ (12.7±5.4). No other statistically significance differences were found ($p>0.05$).

Discussion

The purpose of our study was to investigate lower extremity biomechanics of professional dancers and non-dancers during multidirectional single-leg landings. We hypothesized that 1) professional dancers would present an upright landing posture at initial contact, as well as higher ankle excursion and PvGRF during landing; and 2) higher knee and lower hip flexion would be exhibited in the lateral jump-landing when compared to the forward or diagonal jump directions. The results of this study support our hypothesis that professional dancers exhibited a more extended landing posture at initial contact, with lower knee and hip flexion, higher plantarflexion angle and presented higher ankle excursion during landing. Further, each jump-landing direction elicited distinct lower extremity biomechanical responses, suggesting that sagittal and frontal planes displacements influence overall single-leg landing strategies.

Lower extremity alignment during landing

Professional dancers landed with an everted ankle position in the forward jump-landing compared to the diagonal and lateral non-dancers jump-landings. This suggests that professional dancers avoid a "sickling" pattern (35), maintaining an everted ankle position while plantarflexed. Comparatively to non-dancers, this mechanism is most likely associated to professional dancers' specific and extensive ankle training to maintain the alignment of the leg, ankle and foot. Thereby, they avoid a common mechanism of ankle sprain (ankle inversion with plantarflexion), perhaps due to an increased peroneus muscle activity. During the forward jump-landing, while dancers landed in an everted ankle position, they concurrently exhibited an adducted knee angle. Accordingly, dancers' biomechanical strategy emphasizes the LE alignment. Therefore, the typical "dynamic valgus" mechanism (36) during landing was not observed in the professional dancers. This suggests that the relation between knee and ankle in our study may be a potential protective factor to maintain appropriate lower extremity alignment during landing.

A primary goal during landing is to stabilize and maintain the center of mass within the base of support; the various jump directions influence the strategies adopted to control the landing (25). In our study, professional dancers presented lower hip flexion during the lateral jump-landing compared to the forward jump, as well as when compared to the three jump-landing directions of non-dancers. The lower hip flexion posits that professional dancers recruit the pelvic and hip stabilizers due to the increased lateral displacement. Consequently, dancers tend to prevent the "dynamic knee valgus", associated with excessive hip adduction and internal rotation, probably due to the higher posterolateral pelvis and hip musculature demands (15). Analogous to previous literature, it appears that during single leg-landings, the impact forces were mainly attenuated distally (i.e., knee and ankle joints), due to the smaller base of support, decreasing the work demands on the proximal joints (i.e., hip joint) (27). It is likely that professional dancers' pelvis and hip joint musculature was primarily activated to maintain the postural control of the proximal segments, while the distal segments mainly attenuated the impact forces.

Dance training may influence landing pattern

At initial contact, professional dancers landed in a more extended posture compared to non-dancers. This upright position has been associated with higher PvGRF, which increases the load on non-contractile structures of proximal LE joints (27). Conversely, when using an active trunk flexion strategy during landing, decreased PvGRF, peak knee extensor moment and quadriceps activation and higher hip and knee flexion angles, and hip extensor moment were reported (19, 31, 37). The magnitudes of the attained hip and knee flexion (51º and 82º, respectively) were suggested to be protective of joint loads (19, 37). However, it is plausible that such magnitudes have a deleterious effect on performance in most athletic populations, due to the excessive active trunk flexion (i.e., 96º of active trunk flexion). It is worth noting that in our study, professional dancers exhibited a more erect position at initial contact compared to non-dancers, but no difference in the magnitude of PvGRF was observed; this is most likely due to the higher distal joints excursion, which possibly contributes to offset the load on the passive structures while dissipating the external forces (13, 17, 18). This suggests that the knee and ankle joints can contribute nearly equally to the total shock absorption, with the hip joint contributing minimally (18). Professional dancers appropriately take advantage of their extensive training of the *demi-plié* movement to accomplish this landing strategy. Notably, the ankle, paramount to the aesthetics and technique in dance, plays an important role to adequately mitigate the landing forces (27). The significantly higher excursion and longer time to peak knee flexion probably increases the eccentric action of the plantarflexor muscles which has been shown to decrease PvGRF (14, 17).

Professional dancers presented a higher knee adduction angle, lower hip abduction angle and higher hip abductor moment during landing when compared to non-dancers. Our findings are similar to Orishimo and colleagues that reported dancers (males and females) displayed higher knee adduction angle compared to other athletes (21). It is reasonable to suggest that professional dancers had higher medial-lateral neuromuscular control of the knee joint while increasing lower extremity musculature action rather than relying on the knee ligaments to absorb the landing forces (i.e., less ligament dominance) (36). Further, a neuromuscular link has been observed between the hip and knee with the hip abductors contributing to the neuromuscular control of the knee kinematics during landings (38). While proximal muscular activity was not quantified, based on our hip abductor moment results, it is conceivable that the abductor musculature contributed to minimize the external adductor loads at the hip, and to maintain the neuromuscular control of the knee (38). Overall, when considering both planes of motion (i.e., sagittal and frontal), professional dancers most likely exhibited an improved neuromuscular control than non-dancers during landing. The balanced neuromuscular control observed in professional dancers is probably associated with their extensive practice and a consequence of specific requirements acquired through years, such as aesthetics, elegance, and technique (21).

Largely, professional dancers took advantage of significant ankle and knee range of motion. These structures, particularly the foot-ankle complex, are vastly trained in dancers, such as smoothly land and rolling from the forefoot until a heel touchdown occurs (13). It has been reported that the joint mechanics of the distal joints play an important role on muscle activation patterns to the overall shock absorption strategy (18, 22, 39). It is feasible that dancers' training-specificity leads to automatic pre-planned neuromuscular strategies development that are protective of acute LE injuries during highly skilled activities. Previous research has suggested that initial impact forces can be mediated by automatic strategies through the LE musculature activation patterns (18, 39). The integration of automatic neuromuscular responses must be acquired throughout years of training (40), most likely improving landing patterns and decreasing the deleterious effects of "dynamic knee valgus". These factors combined indicate an efficient control of distal segments, potentially contributing to the lower rate of lower extremity acute injuries in dancers when compared to other populations (28). A comprehensive understanding of the natural biomechanical characteristics between populations, as well as jump directions is essential to enable the development of multifaceted training programs.

Multidirectional jump-landing patterns

Our results demonstrate that the multidirectional jumps have distinctive biomechanical profiles. These differences were more apparent between the forward and lateral directions. We identified a lower plantarflexion, hip flexion, and higher knee flexion, hip abduction angle during the lateral jump when compared to the forward jump. Albeit these differences, during both jumps, participants were in a knee adducted position and presented a similar range of motion at the knee and ankle joints, but a significantly lower hip excursion during the forward direction. These biomechanical patterns suggest distinct neuromuscular demands for each direction, while not necessarily placing participants at increased risk for lower extremity injury. Previous research has primarily employed a drop vertical jump to assess injury risk (10, 13, 17, 19, 20, 31). Though, this task, primarily vertical in nature (33), may not fully represent the demands of various jumps (e.g., vertical vs. horizontal displacements) occurring in several activities (e.g., dance), and the mechanisms of dynamic postural stability (25). Recently, Taylor and colleagues (26) suggested that single-leg landings in different directions (forward and lateral) may better discriminate the biomechanical profile during landing. Injury prevention programs should consider the biomechanical patterns of each direction and adequately integrate them into training regiments.

To maintain the postural control during the lateral jump direction, due to the medio-lateral displacement, higher hip and knee abductor moments were observed at PvGRF likely to provide lateral stabilization of the body and resist the external adduction loads. The multidirectional jumps are commonly performed in activities, such as dance, and sports. Often, injury prevention programs have primarily targeted a single jump direction (e.g., sagittal plane), and it is important to emphasize multidirectional tasks (e.g., multiple planes of motion) due to the interdependency between planes of motion during jump-landing activities. The lateral jump-landing direction has been perceived by the participants as more challenging (25). Anecdotally, we observed, and our participants reported, that the lateral jump-landing direction was perceived as the most difficult to perform. This suggests that the lateral jump-landing direction (e.g. frontal plane) likely constrains the movement degrees of freedom due to the fact that participants may have less perception of the landing area and increase the center of mass oscillation in the frontal plane (medial-lateral control) (16, 25). Therefore, a primary application is that prevention programs should incorporate exercises to develop appropriate frontal-plane neuromuscular responses during jump-landing tasks (25).

Conclusions

Overall, our results suggest that dancers tend to land with a unique biomechanical strategy that may be protective of lower extremity injuries during jump-landings. The professional dancers’ strategies observed in the current study, such as more erect posture and increased forefoot strategy at initial contact, followed by a higher range of motion at the knee and ankle joints, as well as, improved alignment of the distal lower extremity segments, should be considered for implementation in prevention programs. These landing techniques foster adequate lower extremity neuromuscular control with more than likely limited detriment to performance. Dancers’ strategy is achieved over years of training to improve the neuromuscular system in multiple planes of motion (i.e., sagittal and frontal). Further, we also found significant differences among jump-landing directions; specifically, between the forward and lateral jump directions, which are related to the neuromuscular demands of each direction. Given the peculiar ankle pattern and that it is paramount to the aesthetics and technique in dance, future investigation should focus on the intricate foot and ankle complex, to get an insight of the dynamic control mechanism for dissipation of landing forces (e.g., multi-segmented foot model). Future research should also focus on quantifying muscular activation, and the relative muscular contribution and its coordination, through computational modeling of multidirectional jump-landings.

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Conflicts of Interest

There are no conflicts of interest associated with the authors of this study. The results of the present study do not constitute endorsement by the American College of Sports Medicine and are presented without fabrication, falsification, or data manipulation.

References

1. Murphy DF, Connolly DAJ, Beynnon BD. Risk factors for lower extremity injury: a review of the literature. Br J Sports Med. 2002;37:13-29.
2. Caine D, Bergeron G, Goodwin BJ, Thomas J, Caine CG, Steinfeld S, et al. A Survey of Injuries Affecting Pre-Professional Ballet Dancers. J Dance Med Sci. 2016;20(3):115-26.
3. Agel J, Olson DE, Dick R, Arendt EA, Marshall SW, Sikka RS. Descriptive epidemiology of collegiate women's basketball injuries: National Collegiate Athletic Association Injury Surveillance System, 1988-1989 through 2003-2004. Journal of athletic training. 2007;42(2):202-10.
4. Agel J, Rockwood T, Klossner D. Collegiate ACL Injury Rates Across 15 Sports: National Collegiate Athletic Association Injury Surveillance System Data Update (2004-2005 Through 2012-2013). Clin J Sport Med. 2016;26(6):518-23.
5. Ekegren CL, Quested R, Brodrick A. Injuries in pre-professional ballet dancers: Incidence, characteristics and consequences. J Sci Med Sport. 2014;17(3):271-5.
6. Yang J, Tibbetts AS, Covassin T, Cheng G, Nayar S, Heiden E. Epidemiology of overuse and acute injuries among competitive collegiate athletes. Journal of athletic training. 2012;47(2):198-204.
7. Bahr R, Holme I. Risk factors for sports injuries—a methodological approach. Br J Sports Med. 2003;37:384-92.
8. van Mechelen W, Hlobil H, Kemper HCG. Incidence, Severity, Aetiology and Prevention of Sports Injuries. Sports Medicine. 1992;14(2):82-99.
9. Bonci CM. Assessment and Evaluation of Predisposing Factors to Anterior Cruciate Ligament Injury. Journal of Athletic Training. 1999;34(2):155-64.
10. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005;33(4):492-501.
11. Devita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. Med Sci Sports Exerc. 1992;24(1):108-15.

12. Pollard CD, Sigward SM, Powers CM. ACL Injury Prevention Training Results in Modification of Hip and Knee Mechanics During a Drop-Landing Task. Orthop J Sports Med. 2017;5(9):1-7.
13. Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. Am J Sports Med. 2009;37(11):2187-93.
14. Self BP, Paine D. Ankle biomechanics during four landing techniques. Medicine & Science in Sports & Exercise. 2001;33(8):1338-44.
15. Taylor JB, Ford KR, Nguyen AD, Shultz SJ. Biomechanical Comparison of Single- and Double-Leg Jump Landings in the Sagittal and Frontal Plane. Orthop J Sports Med. 2016;4(6):1-9.
16. Sinsurin K, Vachalathiti R, Jalayondeja W, Limroongreungrat W. Different sagittal angles and moments of lower extremity joints during single-leg jump landing among various directions in basketball and voleyball athletes. J Phys Ther Sci. 2013;25(9):1109-13.
17. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clin Biomech (Bristol, Avon). 2003;18(7):662-9.
18. Schot P, Dufek JS. Landing performance part I: kinematic, kinetic, and neuromuscular aspects. Medicine, Exercise, Nutrition and Health. 1993;2:69-83.
19. Blackburn JT, Padua DA. Sagittal-Plane trunk position, landing forces, and quadriceps electromyographic activity. Journal of Athletic Training. 2009;44(2):174-9.
20. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. Clin Biomech (Bristol, Avon). 2010;25(2):142-6.
21. Orishimo KF, Liederbach M, Kremenic IJ, Hagins M, Pappas E. Comparison of landing biomechanics between male and female dancers and athletes, part 1: Influence of sex on risk of anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1082-8.
22. Kovacs I, Tihanyi J, Devita P, Racz L, Barrier J, Hortobagyi T. Foot placement modifies kinematics and kinetics during drop jumping. Medicine & Science in Sports & Exercise. 1999;31(5):708-16.
23. Edwards S, Steele JR, McGhee DE. Does a drop landing represent a whole skill landing and is this moderated by fatigue? Scand J Med Sci Sports. 2010;20(3):516-23.
24. Sinsurin K, Vachalathiti R, Jalayondeja W, Limroongreungrat W. Altered Peak Knee Valgus during jump-landing among various directions in basketball and volleyball athletes. Asian Journal of Sports Medicine. 2013;4(3):195-200.
25. Wikstrom EA, Tillman MD, Schenker SM, Borsa PA. Jump-landing direction influences dynamic postural stability scores. J Sci Med Sport. 2008;11(2):106-11.
26. Taylor JB, Ford KR, Schmitz RJ, Ross SE, Ackerman TA, Shultz SJ. Biomechanical differences in female basketball and soccer players during multi-directional jump landings. J Strength Cond Res. 2017;31(11):3034-45.
27. Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech (Bristol, Avon). 2007;22(6):681-8.
28. Liederbach M, Dilgen FE, Rose DJ. Incidence of anterior cruciate ligament injuries among elite ballet and modern dancers: a 5-year prospective study. Am J Sports Med. 2008;36(9):1779-88.

29. Ambegaonkar JP, Shultz SJ, Perrin DH, Schmitz RJ, Ackerman TA, Schulz MR. Lower body stiffness and muscle activity differences between female dancers and basketball players during drop jumps. Sports Health. 2011;3(1):89-96.
30. Liederbach M, Kremenic IJ, Orishimo KF, Pappas E, Hagins M. Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence of fatigue and implications for anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1089-95.
31. Shimokochi Y, Ambegaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2013;21(4):888-97.
32. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. J Biomech. 1990;23(6):617-21.
33. Fong C, Blackburn JT, Norcross MF, McGrath M, Padua DA. Ankle-dorsiflexion range of motion and landing biomechanics. J Athl Train. 2011;46(1):5-10.
34. Winter DA. Biomechanics and motor control of human movement: John Wiley & Sons; 2009.
35. Jarvis DN, Kulig K. Pointing the foot without sickling: an examination of an ankle movement during jumping. Medical Problems of Performing Artistis. 2015.
36. Hewett TE, Myer GD, Ford KR. Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. Am J Sports Med. 2006;34(2):299-311.
37. Blackburn JT, Padua DA. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. Clin Biomech (Bristol, Avon). 2008;23(3):313-9.
38. Jacobs CA, Uhl TL, Mattacola CG, Shapiro R, Rayens WS. Hip abductor function and lower extremity landing kinematics: sex differences. J Athl Train. 2007;42(1):76-83.
39. Besier TF, Lloyd DG, Ackland TR. Muscle activation strategies at the knee during running and cutting maneuvers. Med Sci Sports Exerc. 2003;35(1):119-27.
40. Simmons RW. Neuromuscular responses of trained ballet dancers to postural perturbations. Int J Neurosci. 2005;115(8):1193-203.

CHAPTER IV

4. Foot Modeling Affects Ankle Sagittal Plane Kinematics during Jump-landing

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Abstract

The foot-ankle complex is a key-element to mitigate impact forces during jump-landing activities. Biomechanical studies commonly model the foot as a single-segment, which can provide different ankle kinematics compared to a multi-segmented model. Also, it can neglect intersegmental kinematics of the foot-ankle joints, such as the hindfoot-tibia, forefoot-hindfoot, and hallux-forefoot joints, that are used during jump-landing activities. The purpose of this short communication was to compare ankle kinematics between a three- and single-segmented foot models, during forward and lateral single-leg jump-landings. Marker trajectories and synchronized ground reaction forces of 30 participants were collected using motion capture and a force plate, during multidirectional single-leg jump-landings. Ankle kinematics were computed using a three- (hindfoot-tibia) and a single-segmented (ankle) foot models, at initial contact (IC), peak vertical ground reaction force (PvGRF) and peak knee flexion (PKF). Repeated measures ANOVAs were conducted ($p<0.05$). The findings of this study showed that during lateral and forward jump-landing directions, the three-segmented foot model exhibited lower hindfoot-tibia dorsiflexion angles (PvGRF and PKF, $p<0.001$) and excursions (sagittal: $p<0.001$; frontal: $p<0.05$) during the weightbearing acceptance phase than the single-segmented model. Overall, the two foot models used provided distinctive sagittal ankle kinematics, with lower magnitudes in the hindfoot-tibia of the three-segmented foot. Furthermore, the three-segmented foot model may provide additional and representative kinematic data of the ankle and foot joints, to better comprehend its function, particularly in populations that the foot-ankle complex plays an important role (e.g., dancers).

Keywords: Multi-segmented foot; Single-segmented foot; Dance; Single-leg landing; Biomechanics.

Introduction

The foot-ankle complex is the link between the lower extremity and the ground (1). Its intricate structure (2-4) enables motion with considerable degree of joint stability (1), and has an essential role to provide dynamic stability during the weightbearing acceptance phase (5). In many athletic and dance activities, the jump-landing movements are fundamental features (6, 7). Jump-landings are part of dancers training throughout years of practice. Dancers are exposed to aesthetic and technical requirements during landing that other athletic populations do not have (e.g., toe-to-heel landing) (7, 8). Furthermore, the landing phase requires dissipation of the kinetic energy generated during the jump phase (6). Thus, the foot-ankle complex not only contributes to initially mitigate the forces transmitted to the body, but also influences the lower extremity biomechanics due to the closed kinetic chain (9-12).

The different foot joints produce a challenging process to define in vivo foot kinematics (13). Recent research has shown clear evidence of how selection of marker data from different bones can influence the kinematics of a segment model (14). In previous landing biomechanical studies, the foot-ankle complex has often been modeled as a single-segment, linking the tibia to the foot (8, 9, 15). However, the bones within each segment of the foot move relative to each other (14). Higher ankle magnitudes have been reported using the single-segmented foot compared to multi-segmented models (16, 17). While the single model is used to quantify foot-ankle kinematics, it neglects the physiological movements between and within the different foot segments (2, 16, 17). The Oxford Foot Model is one of the multi-segmented models that have been employed to investigate the foot-ankle kinematics, particularly the hindfoot-tibia (sagittal, frontal, transverse planes), forefoot-hindfoot (sagittal, frontal, transverse planes) and hallux-forefoot (sagittal plane) joints, providing information of a three-segmented foot (hindfoot, forefoot and hallux) (2, 3). Previous literature has reported that different foot-ankle models can produce distinct ankle joint kinematics, which may lead to inaccuracies in the foot-ankle kinematic outputs (16, 17). Currently, there is limited information on kinematic differences between multi- and single-segmented foot models during single-leg multidirectional jump-landings. Therefore, the purpose of this short communication was to investigate and compare ankle joint kinematics computed by a three- (hindfoot-tibia) and a single-segmented (ankle) foot models. We hypothesized that the hindfoot-tibia would provide lower angles and excursion magnitudes compared to the ankle of the single-segmented model.

Methods

Participants

A total of 30 participants (25.7 ± 5.7 years, 1.69 ± 0.07 m, 61.9 ± 10.1 kg) volunteered to participate in this study, after approval had been obtained from the Institutional Ethical Review Committee (BLINDED FOR REVIEW). Prior to testing, a written informed consent was obtained from each participant. They were required to be between 18 and 40 years old, and physically active with a minimum of 3h/week of physical exercise, such as dance or recreational athletic activities/sports. If participants had a recent history of lower extremity injuries, any pain that would impair the ability to jump, lower extremity surgery within the past five years, or any known neurological/cognitive disorder, they were excluded from the study. The dominant lower extremity was determined as the preferred single-leg landing after performing a countermovement jump (18).

Instrumentation and Experimental Procedures

Kinematic data of the dominant lower extremity were collected (200 Hz) with a 10-camera three-dimensional motion capture system (Opus, Qualisys AB, Gothenburg, Sweden). The ground reaction forces data were recorded (1000 Hz) using a Bertec force plate (FP4060, Bertec Corporation, Columbus, Ohio), and time-synchronized with the kinematic data. Thirty-three retro-reflective markers were placed on selected anatomical landmarks, and a four-marker cluster was used on the thigh of the dominant side. The markers of the foot-ankle complex were placed according to the Oxford Foot Model guidelines (2) [supplementary material]. Four calibration markers (medial femoral condyle and malleolus, upper posterior calcaneus, and first distal metatarsal) were removed prior to the jump-landing trials. Participants stood barefoot and wore a spandex short; and women wore sports bra. Before the static trial, a 5-minute self-directed warm-up was provided.

Participants stood on the non-dominant leg, 70 cm away from the center of the force plate (19), then randomly performed lateral (LJ) and forward (FJ) single-leg jump-landings (19, 20), landed on the dominant leg, in the force plate center. Upon landing, participants immediately transitioned into a maximal vertical jump, followed by a second landing in the force plate center. After being familiarized with the task, participants completed three successful trials for each direction. A rest period of 30 seconds between trials was provided. Trials were excluded if participants lost balance, hopped, stepped off or shifted the dominant foot on the force plate; if they touched the force plate with the non-dominant foot; or if they removed their hands from the hips. For the current study, only the first landing was analyzed.

Data Processing

After markers were manually identified using Qualisys Track Manager (Goteborg, Sweden), biomechanical data were exported to Visual 3D (C-Motion, Inc, Rockville, USA) for data processing. Based on the static trial, a 6-segment kinematic model composed by a pelvis, thigh, shank, hindfoot, forefoot and hallux was created for the multi-segmented model analysis (21); while a 4-segment kinematic model composed by a pelvis, thigh, shank and foot was created for the single-segmented model analysis. Oxford Foot Model kinematic data for the hindfoot-tibia were calculated based on established recommendations (2). The ankle angle of the single-segmented foot model was defined by markers on both malleoli, the calcaneus, the first, second and fifth metatarsals head. The hindfoot-tibia and ankle joints have three degrees of freedom. Knee and ankle joint centers were determined as the midpoints between the medial and lateral epicondyles and malleoli markers, respectively. The hip joint center was estimated using a previously reported regression equation (22).

The kinematic data were calculated during the landing phase: time interval between the initial contact (IC) with the force plate and peak knee flexion (PKF) (8, 23). IC was defined as the point when the vertical ground reaction force exceeded a threshold of 10 N. Peak vertical ground reaction force (PvGRF) was defined as the maximum vertical ground reaction force value during landing. For data analysis, variables were calculated as the mean of the three successful trials of the first landing, for each direction. The dependent variables of interest were sagittal and frontal planes of the hindfoot-tibia (three-segmented foot) and ankle angles (single-segmented model); measured at IC, PvGRF, and PKF. Additionally, we calculated the joint excursions of those joints, computed as the subtraction of the angle between IC and PKF (8). The joint angles were reported in degrees. After conducting a residual analysis on joint kinematics to determine the optimum cut-off frequency (24), a 4th order low-pass Butterworth filter with a 10 Hz cut-off was employed.

Statistical Analyses

Data were analyzed using SPSS (IBM, Chicago, USA). Descriptive statistics and normalcy tests were conducted. Repeated-measures analysis of variance were performed to assess differences between foot models, with each direction analyzed separately. If significant differences were attained, pairwise comparisons with a Bonferroni adjustment were conducted. Statistical significance was set a priori at $p<0.05$.

Results

Descriptive statistics (mean and standard deviation) of the kinematic dependent variables during the lateral and forward jump-landings to compare the three- and single-segmented foot models are presented in Table 2.

Table 2. Descriptive table (mean ± standard deviation) of the multi- (hindfoot-tibia) and single-segmented (ankle) foot models kinematics (angles in degrees) at IC, PvGRF and PKF, in forward and lateral directions.

	IC		**PvGRF**		**PKF**	
	Single (Ankle)	**Multi** (Hindfoot-tibia)	**Single** (Ankle)	**Multi** (Hindfoot-tibia)	**Single** (Ankle)	**Multi** (Hindfoot-tibia)
Forward:						
Plantarflexion (-)	-25.3±8.9	-24.7±7.0	14.6±6.1*	11.0±6.5*	20.0±4.1*	16.1±4.8*
Inversion (+)	-0.08±4.6	-1.6±5.8	-4.8±4.4	-5.2±5.0	-5.7±5.0	-5.2±5.6
Lateral:						
Plantarflexion (-)	-21.1±9.7	-21.9±8.0	22.7±3.9*	17.2±5.2*	25.6±4.2*	19.9±5.2*
Inversion (+)	3.8±4.8	2.2±6.3	-1.4±5.7	-2.0±5.5	-2.5±4.9	-2.6±5.0

There were statistically significant differences regarding ankle kinematics of the two models. During both directions, the three-segmented foot model demonstrated lower dorsiflexion angles at PKF and PvGRF compared to the single-segmented foot (LJ and FJ, $p<0.001$) (Figure 14).

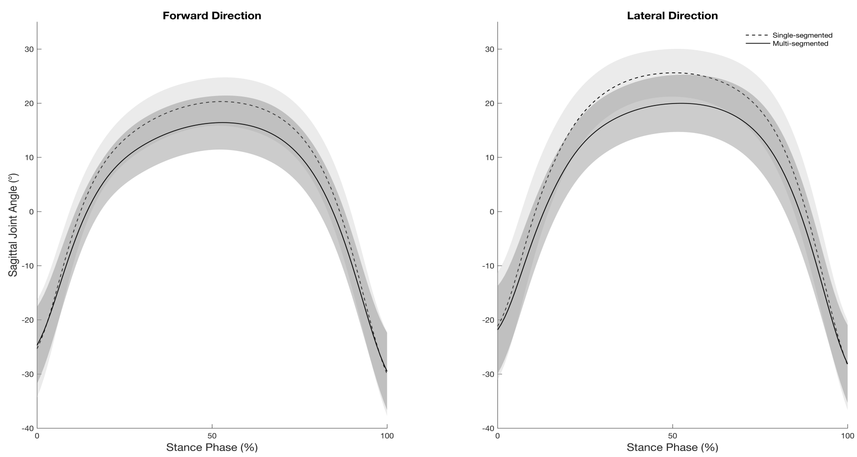

Figure 14. Sagittal plane of the single- (ankle) and multi-segmented (hindfoot-tibia) foot models kinematics (angles in degrees), during forward and lateral jump-landing directions, from initial contact to toe off (stance phase).

Additionally, the three-segmented model had lower sagittal hindfoot-tibia excursion in the lateral (41.8±9.8) and forward (40.8±8.6) directions than the ankle (LJ: 46.7±10.6; FJ: 45.4±10.0) of the single-segmented foot model (LJ and FJ, $p<0.001$). Similarly, in the lateral direction the three-segmented foot also exhibited lower frontal hindfoot-tibia excursion (5.7±4.4) compared to the ankle of the single model (7.1±4.9) ($p<0.05$).

Discussion

The complexity of the foot-ankle anatomy has a noteworthy influence on ankle's biomechanics (1). Previous studies that have investigated the lower extremity landing biomechanics have employed a single-segmented foot model (8, 9, 25). We aimed to investigate ankle kinematics between three- and single-segmented foot models. Our findings demonstrated that the three-segmented foot model exhibited lower hindfoot-tibia dorsiflexion during landing compared to the single-segmented ankle, in both jump-landing directions. Additionally, regardless of jump-landing directions, the three-segmented foot displayed lower hindfoot-tibia excursion in the sagittal plane; while in the frontal plane, only the lateral direction presented statistically significant differences compared to the ankle excursion of the single model.

The talocrural joint is the greatest contributor to plantar and dorsiflexion motions (4). However, this motion is not limited to that joint. The forefoot-hindfoot joint is also a contributor to the overall foot motion (17). Therefore, when computing the foot as a single-segment, the kinematic outputs include not only the hindfoot-tibia joint (talocrural) motion, but also other joints (e.g. forefoot-hindfoot) in a single-segment. Thus, the relative motion between and within the hindfoot-tibia and forefoot-hindfoot is neglected in the single-segmented foot model (2, 17, 26). Subsequently, the single-segmented model simplifies the foot-ankle complex leading to higher kinematic values when compared to multi-segmented foot models, as observed in our study. Previous studies also reported lower sagittal ankle kinematics in the multi-segmented foot model (16, 17). For instance, in the multi-segmented foot model lower dorsiflexion excursion was found in the control (30.19º), chronic ankle instability (27.28º), and coper (28.93º) groups than the single-segmented foot model (43.05º; 37.91º; 40.46º) during side jump-landings (17). Additionally, during walking, the multi-segmented foot model provided lower dorsiflexion excursion (17.6º; 17.1º - normal and flat feet groups respectively), compared to the single-segmented model (25.7º; 24.6º) (16).

During the weightbearing acceptance of the landing phase, landing forces imposed on the body are initially attenuated by the foot-ankle complex, transferring mechanical energy throughout the lower extremity (9, 10). It is suggested that the excursion of the ankle sagittal plane has less influence on kinematics at IC, but greater association with maximum angles and displacements (12). Thus, the ankle weightbearing excursion in this plane has a substantial effect on landing biomechanics (23). The main differences observed in our study were in the sagittal plane, most likely due to being the primary plane where the ankle joint motion occurs and where the body mostly allocates impact forces attenuation (1, 15, 23, 27). Therefore, it is suggested to take into consideration the impact of each foot model on ankle kinematics, as it may affect the entire lower extremity kinetic chain (16).

In conclusion, the two foot models provided distinctive foot-ankle sagittal kinematic magnitudes. The single-segmented foot had considerably higher ankle kinematics than the multi-segmented foot. For future research, the three-segmented foot model should be considered in populations whose foot-ankle joints play a distinguishable role. It provides advanced kinematic data to better comprehend the functional abilities of the intricate foot-ankle joints.

Declaration of Competing Interest: The authors affirm there are no financial and personal relationships with other people or organizations that could inappropriately influence (bias) this work.

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References

1. Brockett CL, Chapman GJ. Biomechanics of the ankle. Orthopaedics and Trauma. 2016;30(3):232-8.
2. Stebbins J, Harrington M, Thompson N, Zavatsky A, Theologis T. Repeatability of a model for measuring multi-segment foot kinematics in children. Gait Posture. 2006;23(4):401-10.
3. Wright CJ, Arnold BL, Coffey TG, Pidcoe PE. Repeatability of the modified Oxford foot model during gait in healthy adults. Gait Posture. 2011;33(1):108-12.
4. Russell JA, Shave RM, Kruse DW, Koutedakis Y, Wyon MA. Ankle and foot contributions to extreme plantar- and dorsiflexion in female ballet dancers. Foot Ankle Int. 2011;32(2):183-8.
5. Towers JD, Deible CT, Golla SK. Foot and Ankle Biomechanics. Seminars in Musculoskeletal Radiology. 2003;7(1):67-74.
6. Tillman MD, Hass CJ, Brunt D, Bennett GR. Jumping and landing techniques in elite women's volleyball. J Sports Sci Med 2004;3:30-6.
7. Liederbach M, Dilgen FE, Rose DJ. Incidence of anterior cruciate ligament injuries among elite ballet and modern dancers: a 5-year prospective study. Am J Sports Med. 2008;36(9):1779-88.
8. Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. Am J Sports Med. 2009;37(11):2187-93.
9. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clinical Biomechanics. 2003;18(7):662-9.
10. Lee J, Song Y, Shin CS. Effect of the sagittal ankle angle at initial contact on energy dissipation in the lower extremity joints during a single-leg landing. Gait Posture. 2018;62:99-104.
11. Donatelli R. Normal Biomechanics of the Foot and Ankle. J Orthop Sports Phys Ther. 1985;7(3):91-5.
12. Hoch MC, Farwell KE, Gaven SL, Weinhandl JT. Weight-Bearing Dorsiflexion Range of Motion and Landing Biomechanics in Individuals With Chronic Ankle Instability. J Athl Train. 2015;50(8):833-9.
13. Okita N, Meyers SA, Challis JH, Sharkey NA. An objective evaluation of a segmented foot model. Gait Posture. 2009;30(1):27-34.
14. Nester CJ, Liu AM, Ward E, Howard D, Cocheba J, Derrick T. Error in the description of foot kinematics due to violation of rigid body assumptions. J Biomech. 2010;43(4):666-72.
15. Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech (Bristol, Avon). 2007;22(6):681-8.
16. Pothrat C, Authier G, Viehweger E, Berton E, Rao G. One- and multi-segment foot models lead to opposite results on ankle joint kinematics during gait: Implications for clinical assessment. Clin Biomech (Bristol, Avon). 2015;30(5):493-9.
17. De Ridder R, Willems T, Vanrenterghem J, Robinson MA, Palmans T, Roosen P. Multi-segment foot landing kinematics in subjects with chronic ankle instability. Clin Biomech (Bristol, Avon). 2015;30(6):585-92.

18. Shimokochi Y, Ambegaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2013;21(4):888-97.
19. Wikstrom EA, Tillman MD, Schenker SM, Borsa PA. Jump-landing direction influences dynamic postural stability scores. J Sci Med Sport. 2008;11(2):106-11.
20. Taylor JB, Ford KR, Nguyen AD, Shultz SJ. Biomechanical Comparison of Single- and Double-Leg Jump Landings in the Sagittal and Frontal Plane. Orthop J Sports Med. 2016;4(6):1-9.
21. Carson MC, Harrington ME, Thompson N, O'Connor JJ, Theologis TN. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. Journal of Biomechanics. 2001;34:1299-307.
22. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. J Biomech. 1990;23(6):617-21.
23. Fong C, Blackburn JT, Norcross MF, McGrath M, Padua DA. Ankle-dorsiflexion range of motion and landing biomechanics. J Athl Train. 2011;46(1):5-10.
24. Winter DA. Biomechanics and motor control of human movement: John Wiley & Sons; 2009.
25. Orishimo KF, Liederbach M, Kremenic IJ, Hagins M, Pappas E. Comparison of landing biomechanics between male and female dancers and athletes, part 1: Influence of sex on risk of anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1082-8.
26. Wright CJ, Arnold BL, Ross SE. Altered Kinematics and Time to Stabilization During Drop-Jump Landings in Individuals With or Without Functional Ankle Instability. J Athl Train. 2016;51(1):5-15.
27. Fong Yan A, Hiller C, Sinclair PJ, Smith RM. Kinematic analysis of sautes in barefoot and shod conditions. J Dance Med Sci. 2014;18(4):149-58.

CHAPTER V

5. Oxford Foot Model Kinematics in Landings: a Comparison between Professional Dancers and Non-dancers

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Abstract

Objectives: Dancers frequently perform jump-landing activities, with the foot-ankle complex playing an essential role to attenuate the landing forces. However, scarce research has been conducted in professional dancers multi-segmented foot in landings. The aim of this study was to compare the multi-segmented foot kinematics between professional dancers and non-dancers, during forward and lateral single-leg jump-landings.

Design: Descriptive group comparison.

Methods: Marker trajectories and synchronized ground reaction forces of 15 professional dancers and 15 non-dancers were collected using motion capture and a force plate, during multidirectional single-leg jump-landings. Sagittal and frontal hindfoot-tibia, forefoot-hindfoot, and hallux-forefoot kinematics of the multi-segmented foot model were computed at initial contact, peak vertical ground reaction force and peak knee flexion. Repeated measures ANOVAs were conducted ($p<0.05$).

Results: Professional dancers landed with higher hindfoot-tibia and forefoot-hindfoot plantarflexion angles at initial contact ($p<0.001$), and hindfoot-tibia dorsiflexion angles at peak vertical ground reaction force and peak knee flexion ($p<0.001$). Also, dancers exhibited higher sagittal hindfoot-tibia and forefoot-hindfoot excursions than non-dancers ($p<0.001$). No statistically significant differences were found in the frontal plane.

Conclusions: The multi-segmented foot allows a comprehensive kinematic analysis of the different foot joints. In jump-landings, professional dancers higher hindfoot-tibia, and forefoot-hindfoot plantarflexion at initial contact, compared to non-dancers, contributed to a subsequent higher foot joints excursion. This pattern is commonly linked to a better shock absorption mechanism in landings.

Keywords: Biomechanics; Dance; Lower extremity; Single-leg landing; Foot-ankle complex.

Practical Implication

- The Oxford Foot Model provides thorough kinematic data to better understand the intricate foot-ankle complex functional abilities.
- In research, the Oxford Foot Model should be considered in populations whose foot-ankle joints play a distinguishable role (e.g. dancers).

- Increased foot-ankle plantarflexion angles at initial contact, and subsequently higher excursions during landing, as observed in professional dancers, might be a strategy to consider to implement in other athletic populations, to potentially improve impact force absorption during landing.

Introduction

Dancers are frequently exposed to jump-landing activities as part of their training,(1) with a ballet dancer performing up to 300 jumps per hour.(2) The jump-landing training in dance, with seemingly effortless landings, produces highly skilled balance ability, and proficiency in jump-landings.(1) Due to aesthetics requirements in dancers performance, obtained over years of training, dancers are used to involve lower extremity alignment and foot work, in a toe-to-heel landing technique, which allows higher ankle plantarflexion angle.(1, 3, 4) The foot-ankle complex, a unique masterpiece of the human body, not only enables motion with high degree of joint stability,(5) but also its intricate structure(6-8) is essential to dancers while performing jump-landings. Additionally, during the landing weightbearing acceptance phase, this structure allows the lower extremity to interact with the ground,(5) influencing the landing biomechanics of the entire lower extremity due to the closed kinetic chain.(9, 10) Thus, it plays an important role to transmit forces between the lower extremity and the ground. Also, it is a key-element that acts to attenuate the landing forces and provides dynamic stability.(11)

The foot-ankle complex has been primarily modeled as a single-segment.(3, 10, 12) While this model is used to measure foot-ankle kinematics, it does not provide information about the relative motion between and within the different foot segments.(6) For instance, in female ballet dancers the talocrural joint is the greater contributor to plantar and dorsiflexion motions (approximately 70% of the overall motion) in weightbearing *en pointé* and *demi-plié* positions, whereas the remaining motion is obtained in other foot joints.(8) Hence, the single model might neglect the internal foot movements, as the foot is comprised more than only the hindfoot-tibia kinematics. It most likely lacks sufficient information regarding the foot-ankle anatomical and physiological functions, providing limited intersegmental angles information of the foot joints.(6, 13, 14) Findings from studies using the single-foot model should be viewed cautiously, because an absence of adaptation at the ankle joint does not imply an absence of adaptation of the whole foot.(15) Additionally, previous assessment of the multi-segmented foot kinematics, during walking and landing, have noted differences that otherwise would not have been detected with a single model,(13, 14) and have reported higher ankle magnitudes using the foot as a rigid segment.(13, 14) Subsequently, different foot models produce distinct ankle joint kinematics, which may lead to inexactitudes in the foot-ankle kinematic outputs.(13)

The foot-ankle biomechanics can provide extensive information regarding its functional abilities and respective influence on the whole lower extremity during landings. Most of the research investigating landing biomechanics have been computing the foot as a single rigid segment (single-segmented foot model) to measure kinematics (Orishimo 2009, Decker 2003, Schmitz 2007). Dancers often rely on the intricate foot-ankle complex while performing jump-landings. In this population, where the foot-ankle structure is a paramount, the multi-segmented foot model should be considered to provide thorough information about the relative motion between and within the different foot segments.(6, 13, 14, 16) The different foot joints create a challenging process to define and represent in vivo foot kinematics using 3D motion capture system. Though, the Oxford Foot Model is one of the multi-segmented foot models that have been applied to investigate the kinematics of the different foot joints.(6, 7) The Oxford Foot Model provides kinematic data of a three-segment foot (hindfoot, forefoot and hallux).(6, 7) Several studies(6, 7, 17) have reported the reliability of this model in adults and children, healthy and pathological populations, in different tasks (e.g. walking, running). However, there is meager scientific information related to the multi-segmented foot in dancers during single-leg multidirectional jump-landings. Therefore, the purpose of this study was to investigate and compare the multi-segmented foot kinematics between professional dancers and non-dancers, during forward and lateral single-leg jump-landings. We hypothesized that professional dancers would exhibit higher plantarflexion and eversion angles and excursions than non-dancers.

Methods

The study was approved by the Institutional Ethical Review Committee (Conselho de Ética da Faculdade de Motricidade Humana, N. 1/2017). Prior to testing, a written informed consent was obtained from each participant. A minimum of 24 participants would be required to achieve 80% statistical power, after an a priori sample size estimation, related to lower extremity biomechanics landings between dancers and athletes,(18, 19) with an effect size of 0.8 and an exploratory alpha level of 0.05. A total of 30 participants were included; 15 professional dancers (26.6 ± 7 years, 1.69 ± 0.1 m, 57.8 ± 9.3 kg), and 15 non-dancers (25.0 ± 5 years, 1.69 ± 0.1 m, 66.0 ± 10.2 kg). They were required to be between 18 and 40 years old. Professional dancers needed to be employed in the professional dance company with a minimum of 10h/week of dancing practice, whereas non-dancers were recreational athletes with a minimum of 3h/week of physical exercise in different sports (basketball, running, surf, gymnasium, swimming football, tennis and boxing). If participants had a recent history of lower extremity injuries, any pain that would impair the ability to jump, lower extremity surgery within the past five years, or any known neurological/cognitive disorder, they were excluded from the study. The dominant lower extremity was determined as the preferred single-leg landing after performing a countermovement jump.(20)

Kinematic data of the dominant lower extremity were collected at 200 Hz with a 10-camera three-dimensional motion capture system (Opus, Qualisys AB, Gothenburg, Sweden). The time-synchronized ground reaction forces data were recorded at 1000 Hz using a Bertec force plate (FP4060, Bertec Corporation, Columbus, Ohio). The same researcher (AMA) obtained the participants' demographics, provided instructions for each task, and placed the markers on the participants. Thirty-three retro-reflective markers were placed on selected anatomical landmarks, and a four-marker cluster was used on the thigh of the dominant side. The markers of the foot-ankle complex were placed according to the Oxford Foot Model guidelines.(6) The retro-reflective markers were secured using double-sided tape, and the marker cluster with a velcro band around the participant's thigh. Four calibration markers (medial knee and malleolus, upper posterior calcaneus, and first distal metatarsal) were removed prior to the performance of the jump-landing trials. Participants stood barefoot and wore a spandex shorts. Additionally, women wore a sports bra during testing. Before the static trial was collected, a 5-minute self-directed warm-up was provided.

Participants stood on the non-dominant leg, 70 cm apart from the center of the force plate,(21) then randomly performed single-leg jump-landings in two directions: lateral and forward(21, 22) and landed on the dominant leg, in the center of the force plate. Upon landing, participants immediately transitioned into a maximal vertical jump, followed by a second landing in the force plate center. Participants were allowed to familiarize themselves with each jump-landing task and completed three successful trials for each direction. A rest period of 30 seconds between each trial was provided. Trials were excluded if participants lost balance, hopped, stepped off or shifted the dominant foot on the force plate; if they touched the force plate with the non-dominant foot; or if they removed their hands from the hips. For the current study, only the first landing of the three successful trials for each direction was analyzed.

A laboratory coordinate system was established with the positive x-axis, y-axis, and z-axis represented anterior, medio-lateral (left side positive), and vertical directions, respectively. After markers were manually identified using Qualisys Track Manager (Goteborg, Sweden), biomechanical data were exported to Visual 3D (C-Motion, Inc, Rockville, USA) for data processing. Anthropometric data including mass and height were used as inputs for each participant. A 6-segment kinematic model composed by pelvis, thigh, shank, hindfoot, forefoot and hallux was created based on the static trial. Oxford Foot Model kinematic data for the hindfoot-tibia, forefoot- hindfoot, and hallux-forefoot joints were calculated based on established recommendations.(6) Knee and ankle joint centers were determined as the midpoints between the medial and lateral epicondyles and malleoli markers, respectively. The hip joint center was estimated using a previously reported regression equation.(23)

All kinematic data were calculated during the landing phase: time interval between the initial contact with the force plate and peak knee flexion.(3, 24) Initial contact was defined as the point when the vertical ground reaction force exceeded a threshold of 10 N. Peak vertical ground reaction force was defined as the maximum vertical ground reaction force value during the jump-landing task. For data analysis, all variables were calculated as the mean of the three successful trials for each direction. The dependent variables of interest were sagittal and frontal planes of the multi-segmented foot: hindfoot-tibia, forefoot-hindfoot, and hallux-forefoot (only sagittal plane) angles measured at initial contact, peak vertical ground reaction force, and peak knee flexion. Additionally, we calculated the joint excursions, computed as the subtraction of the angle between initial contact and peak knee flexion.(3) The joint angles were reported in degrees. After conducting a residual analysis on joint kinematics to determine the optimum cut-off frequency,(25) a 4th order low-pass Butterworth filter with a 10 Hz cut-off was employed.

Data were analyzed using SPSS (IBM, Chicago, USA). Descriptive statistics and normalcy tests were conducted. Repeated-measures analysis of variance (ANOVA) were performed to assess differences between groups for all dependent variables. If significant differences were attained, pairwise comparisons with a Bonferroni adjustment were conducted. Statistical significance was set a priori at $p<0.05$.

Results

Descriptive statistics (mean and standard deviation) of professional dancers and non-dancers multi-segmented foot kinematic variables during lateral and forward jump-landings are presented in table 3.

Table 3. Descriptive table (mean ± SD) of the multi-segmented foot model (HFTBA: hindfoot-tibia; FFHFA: forefoot-hindfoot; HXFFA: hallux-forefoot) kinematics (angles in degrees), during lateral and forward directions, at IC, PvGRF and PKF.

	IC			**PvGRF**			**PKF**		
	Lateral Jump	**Forward Jump**	**Total**	**Lateral Jump**	**Forward Jump**	**Total**	**Lateral Jump**	**Forward Jump**	**Total**
HFTBA Plantarflexion (-)									
PD	-27.2±4.1	-29.2±3.5	-28.2±3.8*	19.4±4.2	14.6±6.3	17.0±5.3*	21.9±3.8	18.5±3.6	20.2±3.7*
ND	-16.5±7.4	-20.2±6.8	-18.3±7.0*	14.9±5.2	7.4±4.4	11.1±4.8*	18.0±5.8	13.7±4.7	15.8±5.2*
FFHFA Plantarflexion (-)									
PD	-12.5±7.0	-12.6±6.9	-12.6±6.9*	7.1±5.5	5.8±5.1	6.4±5.3	7.6±5.5	6.7±5.0	7.2±5.3
ND	-2.3±6.4	-3.1±6.1	-2.7±6.3*	9.1±6.1	6.8±5.7	8.0±5.9	9.8±6.3	8.0±5.3	8.9±5.8
HXFFA Plantarflexion (-)									
PD	-2.1±4.9	-11.4±6.2	-6.7±5.6	-0.9±5.9	-7.1±6.5	-4.0±6.2	-0.004±5.7	-6.1±7.0	-3.0±6.4
ND	-0.2±5.4	-7.0±6.7	-3.6±6.1	-2.5±5.8	-6.6±6.8	-4.6±6.3	-2.4±5.9	-5.4±6.4	-3.9±6.2
HFTBA Inversion (+)									
PD	2.4±5.9	-2.5±6.0	-0.01±6.0	-3.3±3.3	-5.5±4.3	-4.4±3.8	-3.8±3.6	-5.3±5.0	-4.5±4.3
ND	1.9±6.9	-0.7±5.6	0.6±6.2	-0.7±6.9	-4.9±5.9	-2.8±6.4	-1.3±6.1	-5.2±6.2	-3.2±6.0
FFHFA Supination (+)									
PD	0.1±6.8	0.6±5.1	0.3±6.0	0.4±6.4	-0.4±4.4	0.04±5.4	0.4±6.2	-0.3±4.4	0.04±5.3
ND	3.6±6.8	1.9±5.5	2.7±6.2	3.4±6.1	2.0±5.2	2.7±5.6	3.5±5.9	2.1±5.0	2.8±5.5

*Statistically significant between professional dancers and non-dancers ($p<0.05$)

At initial contact, peak vertical ground reaction force and peak knee flexion, a statistically significant group main effects between professional dancers and non-dancers multi-segmented foot were attained, in the sagittal plane ($p<0.001$). At initial contact, professional dancers landed with higher hindfoot-tibia and forefoot-hindfoot plantarflexion angles than non-dancers. At peak vertical ground reaction force and peak knee flexion, professional dancers demonstrated higher hindfoot-tibia dorsiflexion compared to non-dancers (Figure 15).

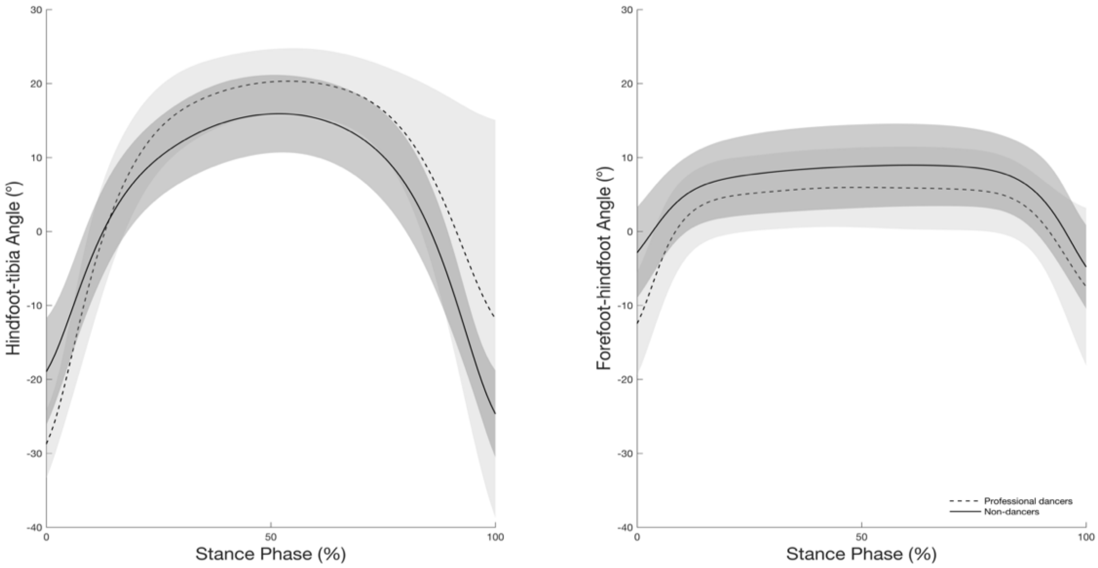

Figure 15. Professional dancers and non-dancers hindfoot-tibia and forefoot-hindfoot angles (in degrees), in the sagittal plane, from initial contact to toe off (stance phase).

Additionally, professional dancers had statistically significant higher hindfoot-tibia (48.4º±3.0) and forefoot-hindfoot (19.8º±4.8) excursions, in the sagittal plane, than non-dancers (34.2º±7.6; 11.6º±3.9) (hindfoot-tibia and forefoot-hindfoot: $p<0.001$) (Figure 16). No other statistically significant findings were observed ($p>0.05$).

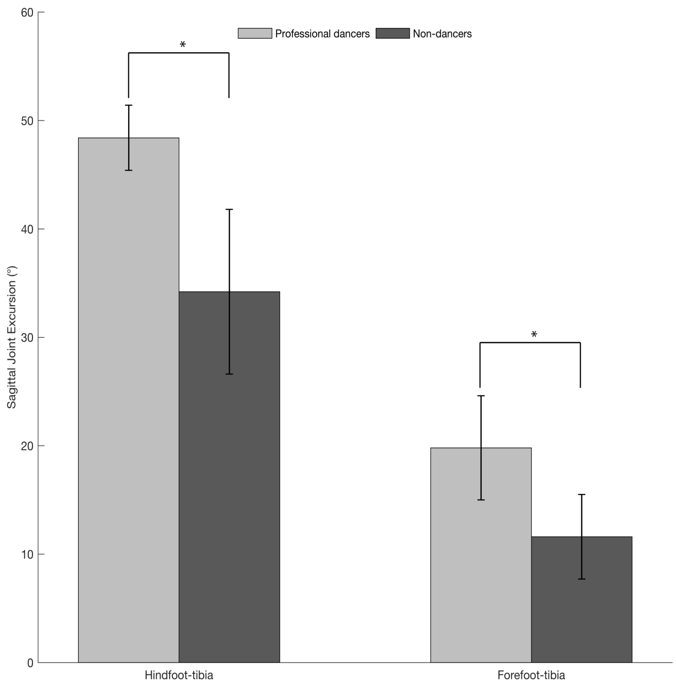

Figure 16. Professional dancers and non-dancers hindfoot-tibia and forefoot-hindfoot excursions (angles in degrees) in the sagittal plane.

*Statistically significant between professional dancers and non-dancers ($p<0.05$).

Discussion

A comprehensive understanding of the foot-ankle biomechanics can provide valuable information related to the function of this intricate structure and its influence on the lower extremity. Considerable research has been conducted to investigate the lower extremity landing biomechanics. Most of the studies have employed a single-segmented foot model,(3, 10, 12) which neglects the relative motion between the different foot segments.(6, 13, 16) This study aimed to investigate the multi-segmented foot kinematics between professional dancers and non-dancers, during forward and lateral single-leg jump-landings. Our findings demonstrated that professional dancers exhibited higher multi-segmented foot kinematic magnitudes in the sagittal plane compared to non-dancers. The findings were observed in the sagittal plane, most likely because it is the primary plane where the ankle joint motion occurs.(5, 12) Furthermore, all participants performed the tasks barefeet, which may potentially influenced the findings, since dancers are more used to perform barefoot, or with slippers or point shoes, whereas non-dancers are more used to sneakers.

During the landing phase, the foot-ankle complex is a key-element that contributes to initially attenuate the forces imposed on the body and to transfer mechanical energy throughout the lower extremity.(10, 26) The complexity of the foot-ankle anatomy has a remarkable influence on the biomechanical performance of the ankle joint.(5) Anatomically, the talocrural joint is the greatest contributor to plantar and dorsiflexion motions.(8) However, motion in the sagittal plane is not limited to the talocrural joint. As previously reported, a substantial amount of plantar and dorsiflexion excursions occur among the joints of the midfoot and forefoot.(8) In dancers, approximately 70% of the total amount of motion to attain the weightbearing *en pointé* and *demi-plié* positions occurs at the talocrural joint; while the remaining 30% is derived from motion between adjacent foot joints.(8) In our study, professional dancers exhibited 48.4º of sagittal hindfoot-tibia and 19.8º of sagittal forefoot-hindfoot excursions, while non-dancers displayed 34.2º and 11.6º, respectively. This shows that in both populations the forefoot-hindfoot joint is also a contributor to the overall foot motion similar to previous reports.(13) Moreover, adjustments in the forefoot kinematics were observed during landing, to anticipate different surfaces inclination.(15)

Regardless of the jump-landing directions, professional dancers exhibited higher hindfoot-tibia and forefoot-hindfoot plantarflexion at initial contact, and higher hindfoot-tibia dorsiflexion at peak vertical ground reaction force and peak knee flexion than non-dancers. It seems that dancers had a specific foot-ankle strategy in landings which can influence their lower extremity motion patterns due to the closed kinetic chain. This is probably an outcome of professional dancers training-specificity. Thus, the experience/background of the participants may contributed to the observed findings of this study. Since young age, dancers training promotes a specific motor organization patterns to adjust the performance requirements and technique, which provides greater mobility to the different joints of the body, including the segments of the foot. The foot-ankle complex function is crucial to dancers' ability to perform, requiring flexibility and strength through the foot and ankle.(27) Also, in landing activities, the foot-ankle complex is a considerably well-trained structure in dancers to enable a smooth landing transition from the toes until the heel contacts the ground.(3)

Increased plantarflexion angle at initial contact has been reported to provide a mechanical advantage due to a subsequent increased ankle excursion during landing, which is established as an important contributor to dissipate the impact forces(10, 26, 28) Furthermore, in single-leg landings the ankle angle at initial contact determines the redistribution of energy dissipation between the ankle and the hip joints.(26) In our study, this landing strategy was observed in the professional dancers, whereas non-dancers had a lower plantarflexion at initial contact that was accompanied with less hindfoot-tibia and forefoot-hindfoot excursions. This strategy may lead to a less efficient landing strategy to attenuate the ground reaction forces.(9, 26) Even more because a higher plantarflexion at initial contact during single-leg landings is associated with increased ankle energy dissipation.(26) Noteworthily, a model-driven study reported that an increased plantarflexion angle at initial contact might increase the susceptibility of ankle sprain occurrence.(29) Contrastingly, it has also been reported that individuals with functional or chronic ankle instability presented lower ankle plantarflexion at initial contact when compared to coper or control groups;(13, 16) this suggests a link to a lower dynamic stability pattern due to the decreased time over which the foot-ankle complex has to absorb the impact forces.(16) Professional dancers had a higher plantarflexion at initial contact, which has been speculated as increased risk for ankle sprains.(29) Moreover, their increased excursion in the sagittal plane while landing may counteract the suggested detrimental ankle position, and assist in the dissipation of ground contact forces as a protective mechanism.(10, 16)

There is strong evidence suggesting that a decreased weightbearing ankle sagittal excursion modifies lower extremity mechanics and may be linked to injuries due to a compensatory mechanical pattern in the lower extremity.(30) Individuals with chronic ankle instability had higher peak vertical ground reaction force, suggesting a negative association with the injury risks due to the limited capacity to absorb loading rates.(13) Thus, greater weightbearing ankle sagittal excursion is associated with greater knee flexion excursion and lower ground reaction forces during landing.(24) A higher plantarflexion at initial contact may enable increased ankle joint excursion for the shock absorption mechanism in single-leg landings.(26) Non-dancer's lower plantarflexion at initial contact and hindfoot-tibia excursion may contribute to the lower knee sagittal excursion during landing, and quicker time to achieve peak knee flexion (end of the landing phase) compared to professional dancers.(4) Furthermore, it has been observed that the lower extremity segment orientation and muscular pre-activation of the musculoskeletal system before initial contact influences the landing performance.(15) If training the landing phase in a highly plantarflexed position at initial contact, athletes may maximize the available dorsiflexion excursion in the weightbearing phase. Thus, the awareness of using more plantarflexion and higher excursion of the foot-ankle complex might be considered to be translated to other athletic populations to plausibly take a better advantage of the landing technique. This is pertinent as athletes may not use the full potential energy-absorbing characteristics of the ankle plantarflexors,(28) which is considered as a protective mechanism to mitigate the landing forces, and minimize the deleterious consequences of the impact forces in jump-landing activities, without compromising performance.(10, 16)

Conclusion

Overall, it was notable that professional dancers landing pattern had higher hindfoot-tibia and forefoot-hindfoot plantarflexion and consequently higher excursion, which may play as a contributor to mitigate the landing forces. This is most likely linked to dancers training-specificity. For future studies, the multi-segmented foot model should be considered for research to provide comprehensive foot kinematic data of the foot-ankle function. Additionally, further research is needed to investigate dancers' lower extremity relative muscular contribution to shock absorption by muscular energy dissipation in multidirectional jump-landings.

References

1. Liederbach M, Dilgen FE, Rose DJ. Incidence of anterior cruciate ligament injuries among elite ballet and modern dancers: a 5-year prospective study. Am J Sports Med. 2008;36(9):1779-88.

2. Wyon MA, Twitchett E, Angioi M, Clarke F, Metsios G, Koutedakis Y. Time motion and video analysis of classical ballet and contemporary dance performance. Int J Sports Med. 2011;32(11):851-5.

3. Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. Am J Sports Med. 2009;37(11):2187-93.

4. Azevedo AM, Oliveira R, Vaz JR, Cortes N. Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings. Med Sci Sports Exerc. 2019;51(3):539-47.

5. Brockett CL, Chapman GJ. Biomechanics of the ankle. Orthopaedics and Trauma. 2016;30(3):232-8.

6. Stebbins J, Harrington M, Thompson N, Zavatsky A, Theologis T. Repeatability of a model for measuring multi-segment foot kinematics in children. Gait Posture. 2006;23(4):401-10.

7. Wright CJ, Arnold BL, Coffey TG, Pidcoe PE. Repeatability of the modified Oxford foot model during gait in healthy adults. Gait Posture. 2011;33(1):108-12.

8. Russell JA, Shave RM, Kruse DW, Koutedakis Y, Wyon MA. Ankle and foot contributions to extreme plantar- and dorsiflexion in female ballet dancers. Foot Ankle Int. 2011;32(2):183-8.

9. Hoch MC, Farwell KE, Gaven SL, Weinhandl JT. Weight-Bearing Dorsiflexion Range of Motion and Landing Biomechanics in Individuals With Chronic Ankle Instability. J Athl Train. 2015;50(8):833-9.

10. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clinical Biomechanics. 2003;18(7):662-9.

11. Towers JD, Deible CT, Golla SK. Foot and Ankle Biomechanics. Seminars in Musculoskeletal Radiology. 2003;7(1):67-74.

12. Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech (Bristol, Avon). 2007;22(6):681-8.

13. De Ridder R, Willems T, Vanrenterghem J, Robinson MA, Palmans T, Roosen P. Multi-segment foot landing kinematics in subjects with chronic ankle instability. Clin Biomech (Bristol, Avon). 2015;30(6):585-92.
14. Pothrat C, Authier G, Viehweger E, Berton E, Rao G. One- and multi-segment foot models lead to opposite results on ankle joint kinematics during gait: Implications for clinical assessment. Clin Biomech (Bristol, Avon). 2015;30(5):493-9.
15. Morey-Klapsing G, Arampatzis A, Bruggemann G. Evidence of Proactive Forefoot Control During Landings on Inclined Surfaces. Journal of Motor Behavior. 2007;39(2):89-102.
16. Wright CJ, Arnold BL, Ross SE. Altered Kinematics and Time to Stabilization During Drop-Jump Landings in Individuals With or Without Functional Ankle Instability. J Athl Train. 2016;51(1):5-15.
17. Milner CE, Brindle RA. Reliability and minimal detectable difference in multisegment foot kinematics during shod walking and running. Gait Posture. 2016;43:192-7.
18. Ambegaonkar JP, Shultz SJ, Perrin DH, Schmitz RJ, Ackerman TA, Schulz MR. Lower body stiffness and muscle activity differences between female dancers and basketball players during drop jumps. Sports Health. 2011;3(1):89-96.
19. Liederbach M, Kremenic IJ, Orishimo KF, Pappas E, Hagins M. Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence of fatigue and implications for anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1089-95.
20. Shimokochi Y, Ambegaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2013;21(4):888-97.
21. Wikstrom EA, Tillman MD, Schenker SM, Borsa PA. Jump-landing direction influences dynamic postural stability scores. J Sci Med Sport. 2008;11(2):106-11.
22. Taylor JB, Ford KR, Nguyen AD, Shultz SJ. Biomechanical Comparison of Single- and Double-Leg Jump Landings in the Sagittal and Frontal Plane. Orthop J Sports Med. 2016;4(6):1-9.
23. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. J Biomech. 1990;23(6):617-21.
24. Fong C, Blackburn JT, Norcross MF, McGrath M, Padua DA. Ankle-dorsiflexion range of motion and landing biomechanics. J Athl Train. 2011;46(1):5-10.

25. Winter DA. Biomechanics and motor control of human movement: John Wiley & Sons; 2009.
26. Lee J, Song Y, Shin CS. Effect of the sagittal ankle angle at initial contact on energy dissipation in the lower extremity joints during a single-leg landing. Gait Posture. 2018;62:99-104.
27. Russell JA, McEwan IM, Koutedakis Y, Wyon M. Clinical anatomy and biomechanics of the ankle in dance. J Dance Med Sci. 2008;12(3):75-82.
28. Self BP, Paine D. Ankle biomechanics during four landing techniques. Medicine & Science in Sports & Exercise. 2001;33(8):1338-44.
29. Wright IC, Neptune RR, van den Bogert AJ, Nigg BM. The infuence of foot positioning on ankle sprains. J Biomechanics. 2000;33:513-9.
30. Mason-Mackay AR, Whatman C, Reid D. The effect of reduced ankle dorsiflexion on lower extremity mechanics during landing: A systematic review. J Sci Med Sport. 2017;20(5):451-8.

CHAPTER VI

6. Effect of Two Different Pose Estimation Approaches on Lower Extremity Biomechanics in Professional Dancers

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Abstract

Different algorithms can be used to estimate the pose of musculoskeletal models in biomechanical studies. Visual 3D uses segment optimization whereas OpenSim uses global optimization. Thus, our purpose was to study whether the two approaches would influence the estimation of lower extremity biomechanical parameters. Marker trajectories and ground reaction forces of 6 professional dancers were collected during a single-leg forward jump-landing. The same data set was processed using both approaches. Our findings suggested that the sagittal knee and ankle angles and moments were highly comparable between the two approaches. The ankle sagittal angle and moment showed the lowest offset. On the other hand, the choice of a kinematic model was likely to affect the hip, more evident in the frontal and transverse planes. This may be due to different factors such as the pelvis and femur positions or larger amount of soft tissue in the thigh.

Introduction

In biomechanical studies, different modeling approaches have been used to investigate lower extremity (LE) kinematic and kinetic parameters. Each model includes specific joint constraints imposed to the relative motion of adjacent body segments resulting in different overall full-leg motion (1). Segment or global optimization approaches can be used to compute kinematics of a musculoskeletal (MSK) model during a dynamic activity. The primary distinction is in the pose (position and orientation) estimation algorithms used. Visual 3D (C-Motion, Inc. Rockville, USA) and OpenSim (2) are two commonly used software to examine biomechanics of human movement.

Visual 3D uses segment optimization to compute segment position and orientation creating a model through individual body segments. Each body segment can apply forces to adjacent segments. For this approach, a static trial is first used to define participant-specific joint centers, followed by segments, joint axes, and segmental coordinate systems. The location of participant-specific joint centers are stored virtually and relative to external kinematic markers placed over the skin of the participant. During dynamic trials, participant-specific joint centers are derived directly from these externally skin-mounted kinematic markers, and joint kinematics are calculated (3, 4). In this approach, each joint is considered to have 6 degrees of freedom (DOF) to describe its kinematics. The segments can move freely about the 3 rotational and 3 translational

DOF, which prevents the propagation of errors from one segment to another (4). However, errors in joint angle estimates can be produced, introducing joint displacements errors and/or changes in segment lengths (5). Additionally, segments are independent of each other and have unconstraint joints. Such configurations and soft tissue artifact errors can affect segment pose estimation and lead to non-anatomical joint displacements (6, 7). It is a known problem that may affect joint center identification and consequently kinematics estimation. This becomes a critical concern, particularly in dynamic activities (i.e., jump-landings), since it can negatively influence the accuracy and/or reliability of joint kinematics and kinetics estimations (1, 6).

On the other hand, OpenSim employs the global optimization method. It uses anatomical linkage constraints to reduce the effect of errors. This approach is highly dependent on joint kinematic constraints, which are applied to overcome unrealistic joint translations. The optimal fit is determined by considering the entire extremity or body at each frame, instead of each segment independently (5). However, important processes such as the selection of the joint constraint type and the identification of model parameters can still arise and may affect the resultant participant-specific models. The LE kinematics can also be affected by the quality of the experimental and analytical procedures implied in the tasks (1). OpenSim uses a static trial to define a participant-specific MSK model's segment lengths and joint axes. The joint DOFs of the model can be chosen a priori (2). The joint coordinates of the rigid MSK model are optimally adjusted to fit the kinematic marker data to potentially minimize the soft tissue artifact. OpenSim is considered an important software for MSK modeling and simulation of forward dynamics. However, it is still a relatively subjective process because of the highly dependency on operator skill during kinematic modeling (8, 9). Some errors or uncertainties in earlier stages can propagate and affect the kinematic results, leading to noteworthy inaccuracies (10).

Musculoskeletal modeling and simulation of dynamic movement allow the study of athletic performances. The LE has a fundamental role during jump-landing activities to provide balance and absorb impact forces. Dancers often perform single-leg jump-landings with significant horizontal displacement while dancing. Despite the substantial experimental research available on dancers' jump-landing activities, computational modeling has not been adequately explored in this population. To our knowledge there is scarce research to compare LE biomechanical outputs using different modeling approaches in dancers. Therefore, the purpose of this study was to investigate differences in the estimation of the LE joints (hip, knee, ankle) kinematics and kinetics between segmental optimization (Visual 3D) and global optimization (OpenSim) approaches in professional dancers during a single-leg forward jump-landing.

Methods

Participants

For this preliminary study, a convenience sample of 6 professional dancers (28±7 years, 1.67±0.04 m, 54.5±8.5 kg), employed in a professional ballet company was selected from a larger study (11). Participants who had a recent history of LE injuries, any pain that would impair the ability to jump, LE surgery within the past 5 years, or any known neurological/cognitive disorder were excluded. A written informed consent was obtained from all participants prior to testing. All procedures performed in the study were in accordance with the ethical standards of the Institutional Ethical Review Committee and the 1964 Declaration of Helsinki and its later amendments.

Instrumentation & Experimental Procedures

Biomechanical data of the right LE and trunk were collected. Marker trajectories were captured at 200 Hz with a 10-camera three-dimensional motion capture system (Opus, Qualisys AB, Gothenburg, Sweden). The ground reaction forces were recorded at 1000 Hz using a Bertec force plate (FP4060-10; Bertec Corporation, Columbus, Ohio) and time-synchronized with the kinematic data.

The same researcher (AMA) obtained participants' demographics, provided instructions for the task, and placed the markers on selected anatomical landmarks. Thirty-seven retroreflective markers were placed on the right LE and trunk of each participant, using double-sided tape. The calibration markers were removed prior to the performance of the task. After a 5-min self-directed warm-up period, a static trial was collected. Participants started 70 cm apart from the force plate center (12), stood on the left LE, and then performed a single-leg forward jump-landing (FJ), landing on the right LE, in the middle of the force plate. Upon landing, participants immediately performed a maximal vertical jump, followed by a second landing in the force plate center. Three successful FJ trials were recorded. For this preliminary study, one static trial and only the first jump-landing of one successful single-leg FJ trial were analyzed for each subject. Trials were deemed not valid if participants lost balance, hopped, stepped off or shifted the foot on the force plate; if they touched the force plate with the left foot; or if the hands were removed from the hips.

Data processing

Markers were manually identified using QTM (Qualisys Track Manager) and then exported to Visual 3D.

Visual 3D Model: A laboratory coordinate system was established with the positive x-axis, y-axis, and z-axis representing anterior, medio-lateral – left side positive, and vertical directions, respectively. Anthropometric data including mass and height were used as inputs for each subject. Then, a 5-segment kinematic model (trunk, pelvis, thigh, shank and foot of the right side) was created based on the static trial. Knee and ankle joint centers were determined as the midpoints between the medial and lateral epicondyles and malleoli markers, respectively. The hip joint center was estimated using a previously reported regression equation (13). Joint angles and internal moments were calculated using conventional inverse dynamics (14).

OpenSim Model: The laboratory coordinate system of Visual 3D was modified to match the OpenSim coordinate system (x-axis – anterior, y-axis – vertical, z-axis – medio-lateral, right side positive). Then, markers trajectories of the static and FJ trials were exported from Visual 3D into OpenSim. Gait2392 model (23 DOF and 92 muscles) was used (2). Changes including locking the metatarsophalangeal joint and increasing the maximum knee extension to 20º were made on the original model. A joint-constrained three-dimensional MSK model was created with 3 DOF at the trunk and hip joint and 1 DOF at the knee and ankle joints. The MSK model was scaled to match each participant's anthropometry. Then, inverse kinematics analysis was performed to compute joint angles from the marker data. The inverse kinematics tool in OpenSim aligned the modified Gait2392's body segments to closely match the experimental marker positions, minimizing the sum of squares of the error between the measured marker locations and the model's virtual ones (2). Marker errors and RMS errors were attained for each subject. After visual inspection and reference to the OpenSim guidelines, we manually adjusted all markers with maximum error above 0.04 m and maximum RMS error above 0.02 m to meet the recommended marker kinematic accuracy. Then, the inverse dynamics tool was used to compute the joints moments (Fig. 17).

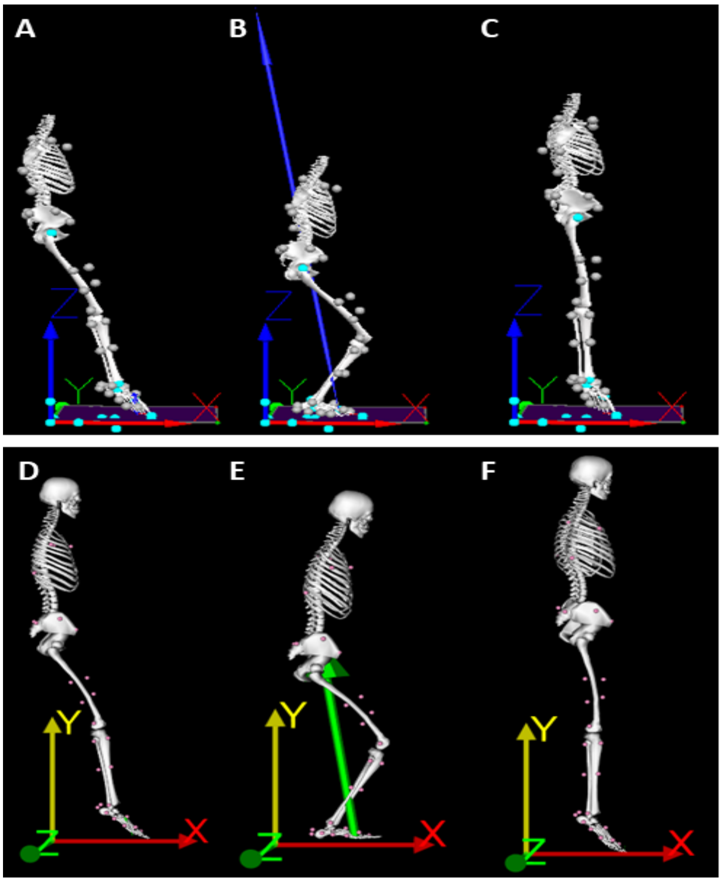

Figure 17. Example of a professional dancer single-leg forward jump-landing at initial contact (A & D); peak knee flexion (B & E); toe off phases (C & F) in Visual 3D (A, B, & C) and OpenSim (D, E, & F).

Visual 3D and *OpenSim*: Initial contact was defined as the point where the vertical ground reaction force surpassed a threshold of 10N (15). Toe off was defined as the instant that the foot stops contact with the force plate. The FJ stance phase was defined as the time period between the initial contact and toe off. The dependent variables of interest were joint angles (º) and internal moments (Nm). Due to the global optimization approach (OpenSim), the kinematic model had joint constraints. Thus, the knee and ankle motions were only reported in the sagittal plane, whereas the hip was reported in the sagittal, frontal and transverse planes.

Matlab: All raw data were exported to Matlab (R2016b). A 4th order Butterworth low pass filter with cutoff frequency at 10 Hz was employed on joint kinematics and kinetics based on the results of residual analysis (14). Subsequently, joint angles and moments were plotted, per joint and for each rotation; also, the offset between the two approaches was computed per joint for each rotation (angles and moments). Lastly, the mean and standard deviation (SD) of the offset were calculated per joint for each rotation.

Results

Overall, the hip, knee and ankle angles and moments in the sagittal plane demonstrated higher similarity in magnitudes and dynamic curve patterns when comparing the two kinematic approaches.

A. Joint kinematics

LE joint angles in the sagittal plane were highly comparable between the two approaches. Furthermore, the knee and ankle sagittal joint angles presented considerably similar magnitudes and dynamic curve patterns for all subjects. The hip sagittal angles showed similar pattern in all subjects, however, with different magnitudes. Visual 3D provided approximately 10º more of hip flexion than OpenSim in all subjects. The hip frontal and transverse planes presented different magnitudes and irregular curve shapes. In the transverse plane, two subjects had an inverse pattern (mirror graphs) of the curves when comparing the two methods, with Visual 3D presenting higher values.

When analyzing mean and SD offset of the LE joint angles, we observed a significantly lower offset in the sagittal plane. The sagittal knee angle had higher dispersion compared with the other joints (Fig. 18).

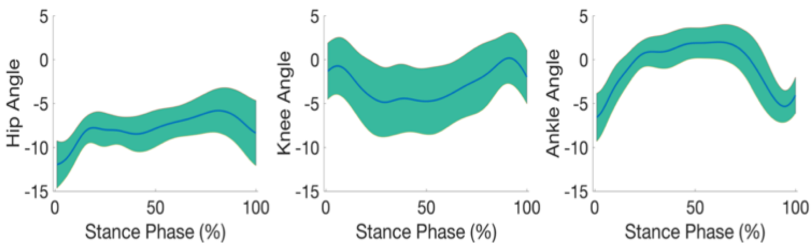

Figure 18. Offset (mean and standard deviation) of 6 professional dancers’ lower extremity joint angles (º) in the sagittal plane during a single-leg forward jump-landing.

B. Joint kinetics

We also observed that the magnitudes and curve patterns in the LE joints were similar in the sagittal plane of the joint moment in all subjects during the FJ. Once again, the knee and ankle magnitudes had similar values between OpenSim and Visual 3D. In the hip frontal and transverse planes, magnitudes and curve shapes were significantly different; Visual 3D presented higher values compared to OpenSim, particularly close to the peak knee flexion.

Examining the mean and SD offset, we observed that the sagittal ankle moment demonstrated the flattest curve shape. Thus, this joint had lower differences between the two approaches. The sagittal hip moment presented higher dispersion at initial contact, compared to the knee and ankle sagittal moments. Similar to the joint angles, the offset of the sagittal knee moment was the one with a larger dispersion compared to the other two joints. The joint angle offset was less sensitive to the specific kinematic approach, when compared to joint moments offset.

Discussion

This preliminary study used two different modeling approaches (segment and global optimization) to examine and compare the LE kinematic and kinetic parameters during a single-leg FJ in professional dancers. The outputs were obtained through the same set of data but computed independently by each method.

Our findings showed that the sagittal plane variables were less affected by the models' characteristics (Fig. 19). This suggests that the observed magnitudes and curve patterns were more consistent for all subjects between the two approaches, particularly in the knee and ankle. Consequently, the sagittal plane was less sensitive to the chosen kinematic model, which is similar to those observed in other tasks (7).

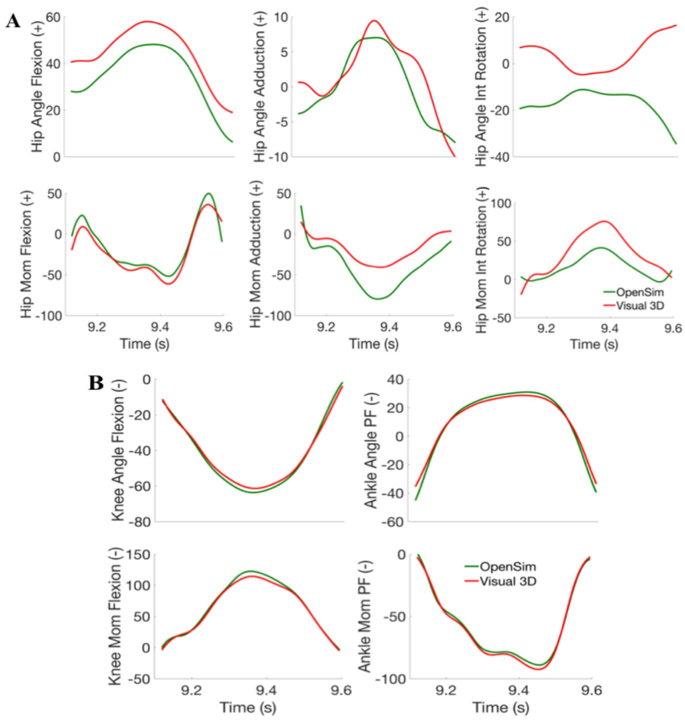

Figure 19. Example of a professional dancer: (A) hip angles (º) and moments (Nm) and (B) knee and ankle angles (º) and moments (Nm), from initial contact to toe off, during a single-leg forward jump-landing.

The hip sagittal angle showed similar shape but different magnitudes. That difference was constant during the stance phase; Visual 3D provided approximately 10º greater of hip flexion during the FJ than OpenSim for all subjects. It is interesting to emphasize that the sagittal plane was the one that had lower dispersion, which is the same plane of the jump-landing task performed (forward direction). Future research should analyze the same task in the frontal plane (lateral direction) to study whether the direction of the task has significant influence on the biomechanical outputs. Hip angles and moments in the frontal and transverse planes showed lower similarity in magnitudes and curve shapes between the two approaches. In this study, those planes demonstrated significantly higher sensitivity to the kinematic approaches than the sagittal plane. This may be due to factors such as the pelvis position and femoral rotation (7). In addition, the thigh has larger amount of soft tissue artifact compared to the leg. Therefore, for hip kinematics estimation, comparative assessment should be carried out with respect to the propagation of the soft tissue artefacts (16). Also, the thigh cluster used may have not been in the exactly same position for all subjects, which could influence our findings.

Previous literature reported that anatomical constraint models seem to provide results with more physiological motion at the joints (1). Yet, our study showed that for the sagittal plane, the results were comparable between the two approaches. Analyzing the offsets of both, we observed that the joint angles in the sagittal plane were less sensitive to the kinematic model chosen. If the variables of interest are mainly in the sagittal plane, any of the models can be used due to the consistent values between both approaches. Contrastingly, caution must be taken when the purpose is to investigate the hip frontal and/or transverse planes. When comparing biomechanical studies using different modeling approaches focused on the hip joint, researcher should be aware how the modeling approaches may affect the kinematic outputs during dynamic activities. The degree of error resulting from small measurement errors can be quite significantly in the results. The determination of different hip joint center locations can consequently result in different effects on joint kinematics (16) particularly in the frontal and transverse planes. Hip angles and moments can also be affected by the inaccurate hip joint center location. For instance, a propagated error of about -22% in hip moment occurred with a 30 mm error of the joint center location (17).

Lastly, OpenSim provided overall lower kinematic and kinetic magnitudes than Visual 3D. No single best solution is established as definite in the literature (1) and our purpose was not to identify which method produces more accurate joint biomechanics, but rather to provide further knowledge on the effect of different pose estimation methods to support the choice of a suitable approach.

Conclusions

Overall, our results showed that the segment and global optimization approaches produced similar biomechanical estimations in sagittal knee and ankle angle and moment; while the most noticeable differences were observed in the frontal and transverse planes of the hip angle and moment. Therefore, this preliminary study suggests that caution must be taken when interpreting hip angles and moments, using different approaches to create the kinematic model.

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References

1. Leardini A, Belvedere C, Nardini F, Sancisi N, Conconi M, Parenti-Castelli V. Kinematic models of lower limb joints for musculo-skeletal modelling and optimization in gait analysis. J Biomech. 2017;62:77-86.
2. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng. 2007;54(11):1940-50.
3. Besier TF, Lloyd DG, L. CJ, Ackland TR. External loading of the knee joint during running and cutting maneuvers. Med Sci Sports Exerc. 2001;33(7):1168-75.
4. Robinson MA, Donnelly CJ, Tsao J, Vanrenterghem J. Impact of knee modeling approach on indicators and classification of anterior cruciate ligament injury risk. Med Sci Sports Exerc. 2014;46(7):1269-76.
5. Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. J Biomech. 1999;32:129-34.
6. Selbie WS. The evolution of pose estimation algorithms for 3D motion capture data: coping with uncertainty. Rev Port Cien Desp. 2011;11(Suppl 3).
7. Moniz-Pereira V, Cabral S, Carnide F, Veloso AP. Sensitivity of joint kinematics and kinetics to different pose estimation algorithms and joint constraints in the elderly. J Appl Biomech. 2014;30(3):446-60.
8. Lund ME, de Zee M, Andersen MS, Rasmussen J. On validation of multibody musculoskeletal models. Proc Inst Mech Eng H. 2012;226(2):82-94.

9. Trinler U, Baker R. Estimated landmark calibration of biomechanical models for inverse kinematics. Med Eng Phys. 2018;51:79-83.

10. Myers CA, Laz PJ, Shelburne KB, Davidson BS. A probabilistic approach to quantify the impact of uncertainty propagation in musculoskeletal simulations. Ann Biomed Eng. 2015;43(5):1098-111.

11. Azevedo AM, Oliveira R, Vaz JR, Cortes N. Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings. Med Sci Sports Exerc. 2019;51(3):539-47.

12. Wikstrom EA, Tillman MD, Schenker SM, Borsa PA. Jump-landing direction influences dynamic postural stability scores. J Sci Med Sport. 2008;11(2):106-11.

13. Bell AL, Pedersen DR, Brand RA. A comparison of the accuracy of several hip center location prediction methods. J Biomech. 1990;23(6):617-21.

14. Winter DA. Biomechanics and motor control of human movement: John Wiley & Sons; 2009.

15. Taylor JB, Ford KR, Schmitz RJ, Ross SE, Ackerman TA, Shultz SJ. Biomechanical differences in female basketball and soccer players during multi-directional jump landings. J Strength Cond Res. 2017;31(11):3034-45.

16. Cereatti A, Camomilla V, Vannozzi G, Cappozzo A. Propagation of the hip joint centre location error to the estimate of femur vs pelvis orientation using a constrained or an unconstrained approach. J Biomech. 2007;40(6):1228-34.

17. R. S, Leardini A, Cappozzo A, Benedetti MG, Cappello A. Effects of hip joint centre mislocation on gait analysis results. J Biomech. 2000;33:1479-87.

CHAPTER VII

7. The Impact of Marker Adjustment in Musculoskeletal Modeling on Kinematic Parameters

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Current status: **In preparation to be submitted.**

Abstract

Computational modeling and simulation of the musculoskeletal system has been performed to investigate movement performance and quality. Most studies using OpenSim, an open source biomechanical analysis platform, have primarily conducted research in gait analysis, with less focus on jump-landing tasks. During the process of computational modeling, errors and uncertainties in the earlier stages can propagate through the workflow, which may negatively influence the realism of the outputs. The aim of this study was to investigate marker errors after performing inverse kinematics in OpenSim and analyze its influence on the lower extremity joint angles. Marker trajectories and ground reaction forces of 11 participants were recorded during a single-leg forward jump-landing. OpenSim inverse kinematics was performed and the maximum (max) error of each marker was attained. The findings of this study demonstrated that when developing a subject-specific biomechanical model, marker errors resulting from manual marker adjustment have significant influence on the musculoskeletal modeling outputs. Decreased max errors were observed after adjusting all markers needed, and differences before and after adjustments affected the sagittal hip, knee and ankle joint angles. In order to achieve effective marker set optimization during biomechanical modeling, it is essential to measure and examine the errors of all markers to increase subject-specific modeling accuracy.

Keywords: OpenSim; Inverse Kinematics; Musculoskeletal simulation; Computational modeling; Single-leg landings.

Introduction

Musculoskeletal modeling and simulation have been used to study human movement. Those biomechanical studies using computational modeling have been primarily focused on investigating the lower extremity during gait analysis (1-3). The lower extremity also plays an important role in jump-landing to absorb the impact forces applied to the body (4-6). Yet, limited studies using computational modeling on athletic tasks, such as jump-landing exist (7-9). Hence, modeling and simulations are useful tools to help understanding lower extremity joint functions during jump-landing tasks.

OpenSim is an open source computational modeling software that has been widely used in the past decades in biomechanical studies. Researchers use this continuously developed platform to share and improve musculoskeletal models. OpenSim is a powerful tool to analyze quantities that can be difficult to obtain experimentally, such as muscle and joint contact forces (2, 10, 11). The musculoskeletal simulations rely heavily on multibody simulation techniques originally devised for vehicle dynamics and other technical applications (10). However, biological systems introduce additional complexity into the models (10). The first component of any analysis is the development of a musculoskeletal model. Nevertheless, during the process of developing such models, the user-dependent skill is a major concern that is still poorly studied (1, 12). The accuracy level requirement of a model depends on the application and not on the model itself (10, 12). It is therefore up to the user to complete the validation process and assess whether the model is accurate enough to be used for the intended estimation (10). The generic musculoskeletal model needs to be scaled to the participant's anatomical and physiological characteristics. That way, the joint centers and anatomical landmarks are aligned to the specific subject and derived from a static trial. During this phase, caution must be taken when employing a generic model to a specific subject, since errors or uncertainties in earlier stages can negatively propagate and subsequently affect the kinematic outputs (3, 10). These aspects are rarely addressed in the literature (3), and may lead to limited and poor interpretations of musculoskeletal simulations (3, 10). It is essential to assess and report the outcomes of this early stage to assure that the model is an adequate representation of the systems that is simulated (10).

Inverse kinematics is a necessary method in movement analysis (1). In OpenSim inverse kinematics, the quality of the kinematic fit is dependent on how a generic musculoskeletal model and the locations of markers upon it reflect the anthropometry of each individual being measured and the actual positions of those markers during the task (1). It depends not only on the exact placement of the markers on the specific anatomical landmarks, but also on the maximum markers errors measured during inverse kinematics (1). The objective is to decrease the differences between the measured marker locations and the model's virtual markers (13). On one hand, this allows marker set optimization, so the model can closely represent the experimental data (10). But on the other hand, this process introduces subjectivity into the process, since it involves manual adjustments, emphasizing the need of a standardized method so the musculoskeletal model process becomes more efficient and objective. Creating a subject-specific musculoskeletal model is also a time-consuming task (14). Based on the OpenSim recommendations, each max marker error and max Root Mean Square (RMS) error should be below 0.04m and 0.02m, respectively, across all data frames. Additionally, due to the inter-dependency of markers, it is important to identify all markers with max errors above the threshold, and not just the single marker with the max error among all. This allows for concurrent adjustment of markers to equally decrease the error, and potentially be more time efficient. To our knowledge, there is still limited research addressing marker error adjustment and its effect on kinematics. Unquestionably it is important to make sure that this approach is able to produce models that adequately represent the systems of interest (10). Thereby, the aim of this study was to investigate the max marker errors after running OpenSim inverse kinematics, and to compare sagittal hip, knee, and ankle joint angles in three difference processing phases of developing a musculoskeletal computational model: i) before marker adjustment, ii) after single max marker error adjustment, and iii) after all markers with max errors adjusted below the recommended threshold error.

Methods

Participants

For this study, a convenience sample of 11 participants (27±6 years, 1.68±0.05 m, 59.7±9.8 kg) was selected from a larger study (15). Only participants whose dominant lower extremity was the right side were included. They were required to be between 18 and 40 years old, and physically active with a minimum of 3h/week of physical exercise, such as dance or recreational athletic activities/sports. They were excluded if reported a recent history of lower extremity injuries, surgery within the past 5 years, any pain that would impair the ability to jump, or any known neurological/cognitive disorder. Prior to testing, the Institutional Ethical Review Committee approved the study, and a written informed consent was obtained from each participant.

Instrumentation and Experimental Procedures

Marker trajectories and time-synchronized ground reaction forces of the right lower extremity were collected with a 10-camera three-dimensional motion capture system (Opus, Qualisys AB, Gothenburg, Sweden) and a Bertec force plate (FP4060-10; Bertec Corporation, Columbus, Ohio), sampling at 200 Hz and 1000 Hz, respectively. The jump-landing task, its methodology and experimental procedure have been previously described (15). For this study, only the static trial and the first jump-landing of one successful single-leg forward jump-landing were analyzed. In the excluded trials, participants have lost balance, hopped, stepped off or shifted the right foot on the force plate; or touched the force plate with the left foot; or removed the hands from the hips.

Data processing

Markers were manually identified using Qualisys Track Manager (Goteborg, Sweden) and then exported to Visual 3D 6 Professional (C-Motion, Inc, Rockville, USA). The laboratory coordinate system of Visual 3D was modified to match the OpenSim coordinate system, where the x-axis represented the anterior direction, the y-axis the vertical direction, and the z-axis the medio-lateral direction (right side positive). Subsequently, the raw marker trajectory data of the first successful forward trial processed in Visual 3D was exported to OpenSim 3.3 (Simtk.org, Stanford USA). The gait2392 musculoskeletal model (23 degrees of freedom and 92 muscle-tendon units) (13) was used in this study. The following adjustments were conducted to this model: locking the metatarsophalangeal joint and increasing maximum knee extension to 20º. These two modifications were required due to the non-anatomical metatarsophalangeal joint angles obtained in a pilot study, and the knee hyperextension of some participants. A joint constrained three-dimensional musculoskeletal model was created with 3 DOF (degrees of freedom) for the trunk and hip, and 1 DOF for the knee and ankle joints.

A scaled 5-segment kinematic musculoskeletal model composed by trunk, pelvis, and right thigh, shank, and foot was then created for each participant, based on the static trial and using a participant's anthropometric measurements. In this step, the mass properties and the dimensions of the body segments were adjusted to each participant. After scaling the model, dynamic segment poses were tracked through inverse kinematics, to compute lower extremity joint angles from the marker trajectory data. OpenSim inverse kinematics tool was used to aligned the modified Gait2392's body segments of the model, to closely match the experimental markers positions, minimizing the sum of squares of error between the measured marker locations and those of the model's virtual ones (13). The max marker error and max RMS error were attained from all the markers of the specific trial. These data were then exported to Matlab (R2016b) to plot marker trajectories per participant and to identify all markers with max errors that needed to be reduced. After visual inspection and based on OpenSim recommendations, we proceeded in two ways: 1) identifying the single marker that had the max error of all markers and over 0.04m, for each participant; 2) identifying all markers that presented max errors above 0.04m and max RMS error above 0.02m. Those markers were manually adjusted in two different processing phases (single marker adjustment and all markers adjusted), to achieve the recommended marker error threshold (maximum error < 0.04m; maximum RMS error < 0.02m) for each marker. With the optimized marker set, a new model scaling and inverse kinematics were conducted for each situation. This was repeated, if necessary, to achieve the expected marker error threshold for each marker and for each participant. At the end of these procedures, after developing the dynamic musculoskeletal model, data were exported to Matlab (R2016b). Then, the marker trajectories and sagittal joint angles were separately plotted per participant for visual inspection and compared in three different processing phases i) before markers adjustments, ii) after single marker adjustment, and iii) after all markers with errors adjusted.

In this study, stance phase was defined as the time period between initial contact and toe off events. Initial contact was the point where the vertical ground reaction force surpassed 10 N (16); toe off was the instant that the foot stops contact the force plate. The stance phase was subsequently divided into a) landing phase, corresponding to the deceleration phase of a jump-landing, and defined from initial contact to peak knee flexion (17), and b) take-off, defined from peak knee flexion to toe off. The variables of interest for this study were max markers error (m) and joint angles (degrees), in different processing phases (i, ii, iii). For this study the hip, knee and ankle angles were only reported in the sagittal plane.

Results

We observed that marker errors can affect sagittal lower extremity joint angles, during a single-leg forward jump-landing, through visual analysis. This study showed that 4 out of 11 participants require no marker set adjustment, after running inverse kinematics in OpenSim. Analyzing the marker set of the remaining 7, we noticed that 5 had more than one marker with error (between two to four markers) above OpenSim recommended threshold, as the example in Figure 20; whereas 2 participants just had a single marker error above the threshold.

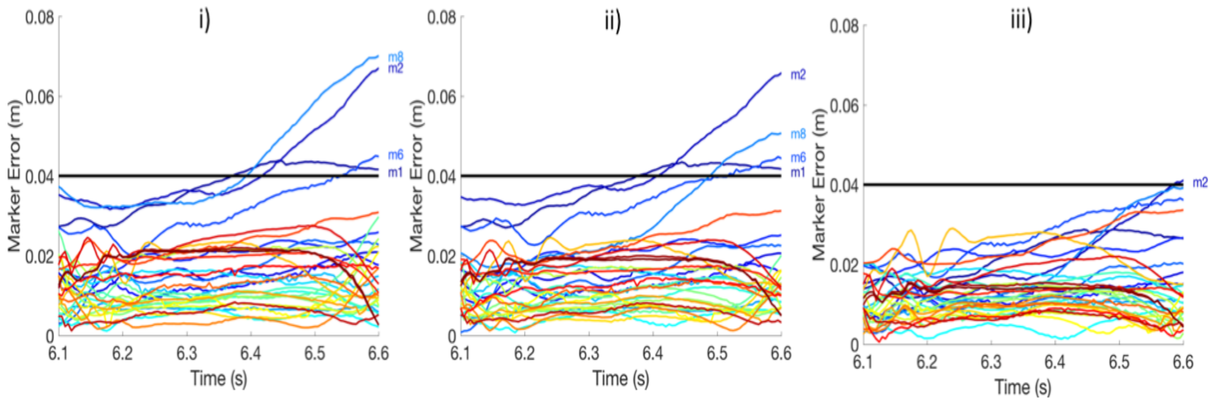

Figure 19. Example of 1 participant markers errors (m) during forward jump-landing stance phase, in the three processing phases: i) before markers adjustment, ii) after single max marker adjusted, iii) after all markers with max errors adjusted. Markers: m8 = right acromion, m2 = left acromion, m6 = left scapula, m1 = C7.

In all 5 participants, comparing phases ii) and iii), there were considerable improvements of marker errors in phase iii). There were 2 participants where in phase ii), the other markers with errors did not improve (just the marker with max error); and there was 1 participant that in that same phase (ii), it resulted in a marker error that was not observed in phase i). The right acromion marker experienced the greatest max fitting error and respective improvement after adjustments in all 7 participants. At the end of the adjustment process of the markers with errors above the recommended threshold, all participants had no marker error on the landing phase; and only 3 had an extremely small error, near the toe off event (end of the take-off phase).

As a preliminary analysis of the sagittal plane of the lower extremity joint angles, we observed that the lower extremity joint angles were affected in the three different processing phases in the 5 participants with more than one marker with max error. In some cases, the joint angles had higher magnitudes in phase iii), as the example depicts in figure 21; whereas in other cases the phase i) displayed higher magnitudes. Moreover, that difference in joint angles was, in some cases, approximately 10º (e.g., ankle of one participant). The hip, knee and ankle joints were always affected by the adjustment of the markers, in the different processing phases, except for 1 participant.

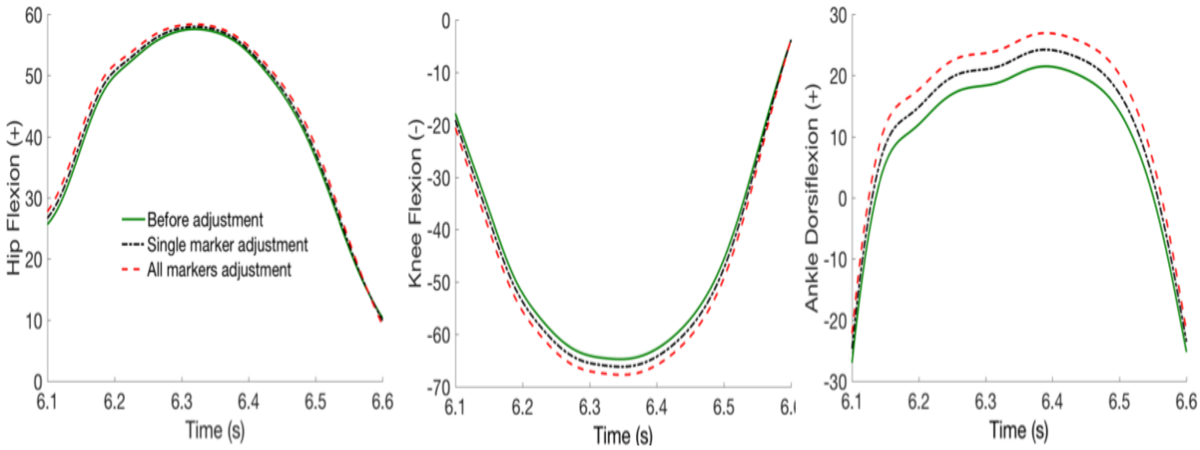

Figure 20. Example of 1 participant lower extremity sagittal joint angles (in degrees) differences during forward jump-landing stance phase, in the three processing phases: i) before markers adjustment (solid green line), ii) after single max marker adjusted (dashed black line), iii) after all markers with max errors adjusted (dashed red line).

Discussion

Computational modeling and simulation is important to further biomechanical analysis (13). However, the subjective manipulation of the user, required to develop a musculoskeletal model, demonstrates that the process is highly dependent on skill and experience (1, 10). The main aim of this study was to investigate the marker set errors after conducting OpenSim inverse kinematics and analyze its effect on lower extremity joint angles during a forward jump-landing task. After analyzing and visually compare the markers error in the three processing phases i), ii), iii), we found that 7 out of 11 participants had markers with error above the recommended values. The majority of the errors were observed in the take-off phase of the jump-landing. This may be possibly explained by sudden and ballistic change in movement direction, suggesting a task-specificity effect. The jump-landing task most likely influenced the trend to present lower or higher marker errors; lower errors were noted during the landing phase, whereas the take-off phase displayed higher number of markers with max error. Besides, the largest and most commonly detected marker with error was the right acromion, once again most likely due to the task-specificity.

All the participants performed a forward-jump landing with the right lower extremity. Additionally, OpenSim models the trunk as a rigid segment; however, even with the hands on the hips required in the methodology of this study, participants could possibly have moved the shoulder upwards. This may hint that mainly during the take-off phase participants used upper body motion, to assist them to propel to the following vertical jump of the task, emphasizing the relationship between segments during dynamic activities. It seems that in some participants, the scapula moved (elevation and upward rotation) during the take-off phase of the forward jump-landing, and more than likely it affected the position of the acromion. Hence, right acromion trajectory during the take-off phase was a noteworthy finding in this study. Other minor marker errors may be due to smaller discrepancies between the model and the participant. Furthermore, after adjustments in all markers with max error (phase iii), we observed an improvement of all marker errors in all participants. Improving just the single max marker error was not enough to optimize the marker set. The marker set optimization was more efficient when considering all markers with error. It is recommended to evaluate and adjust all markers with error that are above the recommended threshold. Additionally, it is also particularly important and should be recommended to acquire experience when manually adjusting the position of the markers, to reduce the error between virtual and experimental markers, and subsequently to closely match the experimental data.

We also observed that even with very similar curve shapes and same directions of rotations, the lower extremity joint angle magnitudes were different between the three processing phases (i, ii, iii). In some cases (e.g., ankle of one participant) the difference achieved approximately 10º. This corroborates that joint angles were sensitive to manual markers adjustments, as they were affected by these modifications during the musculoskeletal modeling process. Moreover, it is unquestionably a subjective process that is influenced by the user's skills, as previously reported (10), and that studies do not explicitly quantify the sensitivity to manual adjustments during the initial musculoskeletal modeling phases (3). The findings of our study also suggest that caution must be taken when interpreting previous results that may have not reported early stage errors and adjustments as those can have propagation consequences to the outputs of modeling and simulation (3, 12).

In conclusion, this study indicated that kinematic marker error in musculoskeletal dynamic simulations are associated with loss in joint kinematics modeling accuracy. The sagittal hip, knee and ankle angles were affected by the manual adjustment of the markers in different processing phases. Moreover, the single max marker error approach was not sufficient to accurately adjust the markers errors. More studies should further examine and comprehend existing uncertainties and limitations during the process of developing a musculoskeletal model into a more standardized process. For instance, previous research has developed a pipeline, proposing a technique for scaling a model and its calibration that needs no intervention of the user. Therefore, it is still important to reduce the subjectivity imposed in this process, when using OpenSim inverse kinematics, due to its influence into the calculation process and subsequently in the results, as observed in this study.

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References

1. Trinler U, Baker R. Estimated landmark calibration of biomechanical models for inverse kinematics. Med Eng Phys. 2018;51:79-83.
2. Lin YC, Dorn TW, Schache AG, Pandy MG. Comparison of different methods for estimating muscle forces in human movement. Proc Inst Mech Eng H. 2012;226(2):103-12.
3. Myers CA, Laz PJ, Shelburne KB, Davidson BS. A probabilistic approach to quantify the impact of uncertainty propagation in musculoskeletal simulations. Ann Biomed Eng. 2015;43(5):1098-111.
4. Schot PK, Dufek JS. Landing performance part I: kinematic, kinetic, and neuromuscular aspects. Medicine, Exercise, Nutrition and Health. 1993;2:69-83.
5. Self BP, Paine D. Ankle biomechanics during four landing techniques. Medicine & Science in Sports & Exercise. 2001;33(8):1338-44.
6. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clinical Biomechanics. 2003;18(7):662-9.
7. DeMers MS, Hicks JL, Delp SL. Preparatory co-activation of the ankle muscles may prevent ankle inversion injuries. J Biomech. 2017;52:17-23.
8. Laughlin WA, Weinhandl JT, Kernozek TW, Cobb SC, Keenan KG, O'Connor KM. The effects of single-leg landing technique on ACL loading. J Biomech. 2011;44(10):1845-51.
9. Mokhtarzadeh H, Yeow CH, Hong Goh JC, Oetomo D, Malekipour F, Lee PV. Contributions of the soleus and gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. J Biomech. 2013;46(11):1913-20.
10. Lund ME, de Zee M, Andersen MS, Rasmussen J. On validation of multibody musculoskeletal models. Proc Inst Mech Eng H. 2012;226(2):82-94.
11. Schellenberg F, Taylor WR, Jonkers I, Lorenzetti S. Robustness of kinematic weighting and scaling concepts for musculoskeletal simulation. Comput Methods Biomech Biomed Engin. 2017;20(7):720-9.
12. Lund ME, Andersen MS, de Zee M, Rasmussen J. Scaling of musculoskeletal models from static and dynamic trials. International Biomechanics. 2015;2(1):1-11.

13. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng. 2007;54(11):1940-50.
14. Valente G, Pitto L, Testi D, Seth A, Delp SL, Stagni R, et al. Are subject-specific musculoskeletal models robust to the uncertainties in parameter identification? PLoS One. 2014;9(11):e112625.
15. Azevedo AM, Oliveira R, Vaz JR, Cortes N. Professional Dancers Distinct Biomechanical Pattern during Multidirectional Landings. Med Sci Sports Exerc. 2019;51(3):539-47.
16. Taylor JB, Ford KR, Nguyen AD, Shultz SJ. Biomechanical Comparison of Single- and Double-Leg Jump Landings in the Sagittal and Frontal Plane. Orthop J Sports Med. 2016;4(6):1-9.
17. Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. Am J Sports Med. 2009;37(11):2187-93.

CHAPTER VIII

8. General Discussion

This PhD thesis was conducted to investigate lower extremity kinematics and kinetics during jump-landings in professional dancers and non-dancers, and provided a significant contribution to advance the body of work in dance science. The current chapter is divided in four sections. The first section provides an overall overview of the main findings according to the research questions of the five scientific papers written during the PhD. Additionally, in the second section, specific limitations of this PhD research are pointed out. Essential recommendations to contribute for future work are presented in the third section. Lastly, in the fourth section the practical implications and its impacts are explained.

8.1 Main findings

One of the imperative methodological decisions in research is which task should be chosen to answer the research question(s). It is undeniable that a better understanding of the movement and forces throughout the lower extremity in high-demand jump-landing tasks is still needed to accurately target the risk factors (25-27, 92). A laboratory task to mimic jump-landings, as more realistic as possible, in different populations was sought in this PhD research. Advanced information was acquired based on prior literature regarding jump-landings (Chapter II – section 2.2). Drop landing or drop jump would not appropriately suit the purposes of this PhD research, and the selected task was single-leg jump-landings in different directions: lateral, diagonal, forward; which professional dancers (ballet and modern *repertoires*) and non-dancers (recreational athletes from different sports) regularly perform.

Additionally, the foot-ankle mechanics is extremely important due to its effect on the proximal joints. Despite the foot being commonly modeled in biomechanics research as a single segment, in this PhD research to study the intricate foot-ankle complex, a multi-segmented foot model was employed (Chapter II – section 2.3), and interesting findings were attained.

Study 1 results suggested that regardless of the group, different directions elicited distinctive lower extremity biomechanical responses, and more prominent between forward and lateral directions, as previously reported (26, 27, 39, 40, 73). That influenced the selection of only the forward and lateral directions for studies 2 and 3. The lateral direction had lower plantarflexion with higher inversion angles at IC, higher dorsiflexion with less eversion angles, higher knee and hip abductor moments compared to other directions. Further, it took longer to achieve PKF and PvGRF, and was perceived by participants as the most challenging direction to perform. This was noted on similar previous studies (39, 73). The lack of visualization of the landing area and the increased center of mass oscillation to acquire lateral stabilization of the body may partially explain this difference. This suggests that jump-landing directions definitely had an impact on biomechanical responses. Professional dancers landed with higher knee adduction angle and hip abductor moment, and lower hip abduction angle than non-dancers. These results further support that professional dancers, through possible increased hip abduction muscles activation, were able to control the proximal joints of the lower extremity; potentially they had increased medio-lateral stabilization of the body, which is in accordance with higher eccentric demands of the medio-lateral hip musculature in the deceleration phase of single-leg landings (27).

The landing pattern of professional dancers reinforced the concept that they prevent the aforementioned sickling mechanism (Chapter II – section 2.3.1), exhibiting great plantarflexion and eversion angles in the forward direction, counteracting the typical anatomic motion of the ankle (kinematically coupled plantarflexion with inversion) (66). This may suggest a protective mechanism for injuries at the ankle (e.g., ankle sprain). Even not quantifying muscle activation, it is plausible that dancers properly activated the peroneus longus and brevis, tibialis posterior and flexor hallucis longus and brevis to achieve the required alignment of the ankle-foot complex (37). This strategy was more distinguished, when also considering the knee joint. In the forward direction, a concomitant knee adduction angle was observed. Accordingly, dancers presented a lower extremity alignment that is protective of the dynamic valgus mechanism (42, 43).

Data presented in this PhD thesis also largely supports that professional dancers may develop skills to achieve proficiency in the landing phase. Some of the findings of study 1 are in line with proposed strategies that are considered as more advantageous to dissipate energy and to reduce loading rates, consequently improving the shock absorption mechanism (19, 21, 41, 53, 67). Particularly, professional dancers landed with an extended posture, lower hip and knee flexion and higher plantarflexion at IC compared to non-dancers (Figure 22); although with no differences in the magnitudes of the PvGRF. In that sense, professional dancers were able to distribute the impact forces over a long period of time, due to their higher knee and ankle excursions and also longer time to achieve PKF when compared to non-dancers, similar to what was found in adolescents females ballet dancers, when performing single-leg hop and stop jump task (93). The body position at IC in jump-landings has been suggested to be a critical determinant of knee injury risk (e.g., anterior cruciate ligament injury), due to its subsequent biomechanics' effects during the landing phase (94). The increased joint excursions in the sagittal plane have been associated to concomitant reduce of the frontal plane joint excursions (14). Altogether, this landing pattern may subsequently contribute to offset the overload on the passive structures and potentially decrease the risk of injuries (19-21). These results imply that dance training technique, when appropriately applied, may be advantageous and considered in other athletic populations.

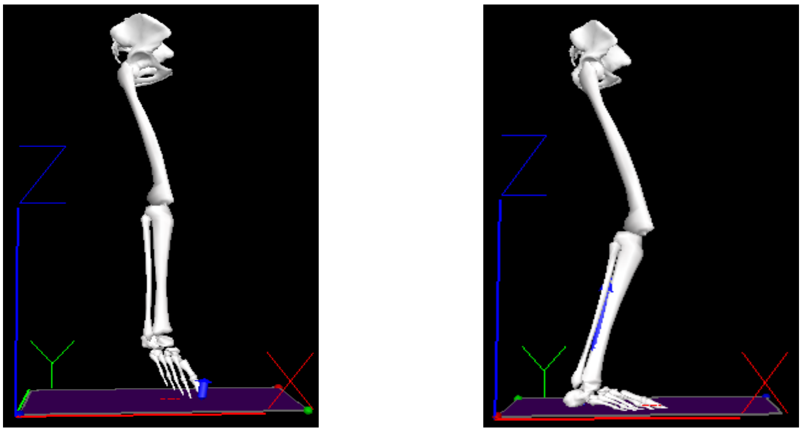

Figure 21. Representation of the sagittal plane dominant lower extremity of a professional dancer (left), and a non-dancer (right) at initial contact during a forward single-leg jump-landing.

The foot-ankle complex is undoubtedly paramount to aesthetics and technique in dance representing an impressively well-trained structure (13, 57). Still, there is a lack of studies in dance science investigating the foot as a multi-segment. Study 1 demonstrated that professional dancers landed with higher ankle plantarflexion at IC than non-dancers, suggesting that ankle angle at IC determines the ankle excursion during single-leg landings (21). Though, to further investigate the foot-ankle complex and advance information, the multi-segmented Oxford Foot Model was adopted. As stated in Chapter

II – section 2.3.2, this model measures hindfoot-tibia, forefoot-hindfoot, and hallux-forefoot joint angles. Study 2 findings provided more detailed data regarding the foot-ankle complex. To start with, the results from that study highlighted the importance of the foot modeling. The multi-segmented foot model displayed lower hindfoot-tibia dorsiflexion angle and excursion, in both jump directions compared to the ankle of the single-segmented model. This was consistent with prior research in jump-landings and walking (80, 81). Additionally, the results from study 3 offered evidence to support two primary facts. Firstly, the hindfoot-tibia joint is not the only contributor to the sagittal plane motion of the foot-ankle complex, even if it has a considerable amount of movement; approximately 63% and 59% of the motion during landing occurred in the hindfoot-tibia of professional dancers and non-dancers, respectively. Similar findings were also observed by other researchers investigating slightly different tasks. Those studies have also used different ways of measurement, such as goniometer (2), and x-ray (1). Furthermore, it was noteworthy to notice in study 3 that at IC, professional dancers had 59% of plantarflexion at the hindfoot-tibia joint, and 27% at the forefoot-hindfoot; while non-dancers had 74% and 11%, respectively. This means that professional dancers unquestionably used more the foot and ankle joints. Secondly, professional dancers landed with higher multi-segmented foot kinematic magnitudes (hindfoot-tibia and forefoot-hindfoot joints) in the sagittal plane than non-dancers. Curiously, there were no differences in the hallux-forefoot joint; which perhaps may be associated to less motion on this joint.

Dancers are trained to plantarflex the foot and ankle in jump-landing movements. Contrastingly, non-dancers most likely use the foot and ankle plantarflexion only to propel the body off the ground, without an effort to maintain the plantarflexion position during the flight phase (66). Study 3 data may partially support this argument. Approximately ¾ of the foot-ankle motion in non-dancers, at IC, occurred in the hindfoot-tibia joint, most likely an action justified by the gravity acting in the foot-ankle joints/segments than active plantarflexors recruitment. Muscle analysis investigation would be valuable to corroborate whether professional dancers present a higher activation of the plantarflexors before and at IC of the landing phase. This could provide support to the concept that the ankle plantarflexors should be considered when developing training strategies to target knee injuries (e.g., anterior cruciate ligament injury) (95). It should be pointed out that the main findings of study 2 and 3 primarily occurred in the sagittal plane. This can be partially explained with the fact that the body mostly allocates impact attenuation in this plane, since it is the primary plane of motion of the foot-ankle joints (20, 74, 85, 96). Undoubtedly the foot-ankle complex is more intricate than a single rigid segment, as confirmed in the findings of this PhD thesis. Therefore, if should be considered in studies with populations whose foot-ankle structure has an important function.

Despite numerous studies investigating biomechanical factors surrounding sport-related jump-landing performance, there is a paucity of focus on dance-related jump-landing. This gap considerably increases when considering the use of computational modeling and simulation to analyze data (e.g., OpenSim). OpenSim has not yet been extensively employed in jump-landing tasks compared to gait analysis. Further, to our knowledge, no study applying computational modeling approaches in dancers' jump-landing have been published.

Musculoskeletal computational modeling and simulation can be a powerful tool in dance research. Simultaneously, this approach presents multiple challenges for its implementation (e.g., user dependent skills). During the process of computational modeling, it is important to test the robustness of the study by evaluating the sensibility of the protocol step by step (31). As such, to be able to fully take advantage of OpenSim, the first step needed in this PhD research regarding computational modeling and simulation was to ensure that this software was providing similar results as Visual 3D, as they use different algorithms to estimate position and orientation of musculoskeletal models. This comparison of software would allow confidence on the interpretation of the results derived from both methods. To accomplish this goal, the same data set was used in both software, and further comparison of lower extremity joint angles and moments was made by visual inspection. The two software used different algorithms for pose (position and orientation) estimation: OpenSim uses global optimization whereas Visual 3D uses segment optimization. The findings of paper 4 demonstrated that the hip joint was the most sensitive to the choice of a kinematic model. This can be partially explained by the following aspects: i) pelvis position and femoral orientation (97), ii) the differences in the amount of soft tissue artifact (90, 98), and iii) joints DOF (89), OpenSim hip model had 3 DOF whereas the knee and ankle had 1 DOF; on the other hand, in Visual 3D all joints had 3 DOF. Contrastingly, the more distal joints of the lower extremity, knee and ankle in the sagittal plane, were less sensitive to the kinematic model. Additionally, OpenSim provided overall lower magnitudes than Visual 3D. In the end, no optimal solution can be established (90), and the research question should drive the software choice.

As presented in chapter II – section 2.4, a subject-specific model is required to be developed that represents each participant's anatomy and physiology. To increase the subject-specific modeling accuracy, through an effective marker set optimization, identification and adjustment of all max markers' error are critical. The findings of paper 5 reported that marker adjustments have an impact on musculoskeletal lower extremity kinematics. The earlier stages to compute a model is therefore extremely important for the entire process and respective accuracy of the outputs. As an example, the scaling and inverse kinematic processes can influence the outputs of loading conditions (99). Also, visualization of models and simulations are critical during the process for interpreting, troubleshooting and communicating results (32). During the process, it is each user's responsibility to ensure that the models and simulations realistically represent the experimental data, and that results from the simulations are truthful (30, 31). Among a variety of potential errors and uncertainties during the process, it still remains unclear how to best achieve robust musculoskeletal modeling analysis and producing minimal errors (99).

Overall, the robustness of this PhD thesis is placed on dancers' unique biomechanical landing profile, led by specific motor organization patterns that they are required to achieve throughout years. This landing pattern may be beneficial to an improved shock absorption mechanism. Scientific basis to establish better strategies to mitigate the landing forces, and subsequently to decrease overload on the passive structures have been proposed over the years (17-19, 21, 42, 65). It seems that the suggested higher flexion of the trunk, aforementioned in Chapter II – section 2.2, to reduce the impact forces (17, 18, 65) would most likely compromise and affect dance and athletic performance, becoming challenging to apply in intervention and prevention programs. Other proposed strategies may be alternative strategies and have more applicability. The findings of studies 1 and 3 further reinforce the argument that dance jump-landing training technique may have some advantages to transfer to other athletic populations. The focus, to develop and foster appropriately injury prevention programs in sports, related to jump-landing tasks, therefore should be on the linkage of these three-key main observations in professional dancers:

1) efficient use of the lower extremity joints; more specifically higher knee and foot-ankle excursions, which is more favorable to mitigate the impact forces;

2) appropriately foot-ankle work. Increased plantarflexion angle at IC increases ankle joint excursion and consequently a muscular strategy to optimize the eccentric ankle plantarflexors activation. Further, increased energy dissipation of the ankle leads to decreased energy dissipation at the hip joint (21). This strategy has been considered to decrease PvGRF and concomitantly attenuate the landing forces (19, 21, 22, 41, 68);

3) association of lower extremity alignment and hip joint kinematics; meaning ankle plantarflexion with eversion and concurrent knee adduction is associated to a higher proximal (hip joint) motion that maintains the medio-lateral neuromuscular control of the pelvis and lower extremity in single-leg jump-landings.

Additionally, the foot-ankle is an extremely important structure in jump-landings as it affects the entire lower extremity biomechanics. Study 2 outcomes highlighted that caution must be taken when modeling the foot due to different ankle angle magnitudes. The multi-segmented foot model should be considered for research in populations whose foot-ankle joints play a noticeable role. It provides further kinematic data to study the foot-ankle joints functional abilities. Lastly, the OpenSim software should be considered to foster research of dancers' muscle analysis and increase the body of work regarding jump-landing activities.

8.2 Limitations

The overall limitations of this PhD thesis are attributed to methodological aspects. This PhD research was undertaken on a sample of professional dancers. They performed a mixed *repertoire* that included classical and contemporary routines. Several differences in postural and movements practices exist between *ballet* and modern dance (11). Furthermore, dancers with less experience, with probably faulty technique and subsequently poorer landing mechanics, were not included as only professional dancers were part of the sample.

There was a substantial effort to apply a task that would mimic, as much as possible, the reality of dancers and non-dancers' performances, as aforementioned in Chapter II – section 2.2.2). The multidirectional single-leg jump-landings were performed in a controlled laboratory environment that may not be entirely indicative of the environment, intensity or demands during dance or athletic performances. Additionally, to minimize movement of the arms and trunk, and prevent the possible enhancement of jumping performance when using arm swing (100), the hands were placed on the hips. The purpose was to quantify as much as possible the biomechanics of the dominant lower extremity. However, we did not investigate unanticipated jump-landing activities, as well as jump height, which should be considered in further work.

Due to the requirement of a multi-segmented foot model, participants needed to perform the jump-landing tasks barefoot. In most landings, professional dancers commonly wear slippers or pointe shoes, and non-dancers typically use sneakers in athletic activities. On the other hand, the use of shoes would have hindered the assessment of certain foot joints motion.

OpenSim has its own limitations. Few examples are: the muscle-tendon’s origin and insertion are based on cadaver studies; it is a subjective process, dependent on users’ skills and experience. With many conceivable errors for relating the observed motion capture to the musculoskeletal model in an inverse kinematics approach, it remains imprecise how best to achieve robust musculoskeletal modeling analysis (99). Thus, it becomes a challenging to use it without “direct help” of an expert that can recognize the steps needed to attain accurate and successful estimations.

8.3 Recommendations for future research

This PhD thesis provided evidence regarding biomechanical landing patterns that highlights: i) several directions in single-leg, ii) a multi-segmented foot model, and iii) application of computational modeling and simulation that have not been previously explored in dance science. The five scientific studies of this PhD thesis characterize the research path done so far. Actually, the PhD is just the beginning of the process. The next steps are already being considered and developed to address questions, particularly related to computational modeling muscle analysis and joint reaction forces in professional dancers. Though, further research in jump-landing is still undoubtedly needed. The recommendations for future research presented in this section are based on scientific and practical relevance.

1) Inadequate shock absorption in jump-landing reduces the ability of the lower extremity joints to mitigate the impact forces, which can easily lead to injuries (25). As reported in this PhD thesis, dancers landing pattern seems to have positive impact in the shock absorption mechanism. Thus, it would be worthy to investigate whether athletes could potentially integrate and adopt some of the aforementioned strategies of dancers landing pattern, and then evaluate its effect on athletes landing patterns. The next step would be towards determining of those strategies the more efficient and easier to implement in athletics activities to be integrated into intervention and injury prevention programs for athletes, and without decreasing their performance.

2) To investigate whether professional dancers can transfer their landing patterns to unanticipated tasks. It would be helpful the use of a markerless system to assess this as well as include in their natural environment.

3) While the influence of the arms in landing activities have been previously described, dance aesthetics does not permit dancers to use forceful arm swings to generate momentum for their jumps (15). It would be valuable to study the influence of the arms in jump-landing tasks and compare with other athletic populations that commonly use them.

4) Muscle and joint forces are difficult to measure experimentally. However, understanding how muscle forces coordinate motion is essential in high performance athletic activities (31). Thus, further research is needed to evaluate dancers’ muscle analysis and associate with kinematic and kinetic results. Also, lower extremity joint reaction forces should be computed using OpenSim, to investigate how the high physical demands placed on dancers’ bodies affect the joints of the lower extremity.

5) Through this PhD research it was established that when developing an OpenSim kinematic model, it is critical to measure and identify all max markers’ error to increase subject-specific modeling accuracy. The computational modeling and simulation field also needs more tools to aid the model validation process – modern algorithms for efficient sensitivity testing and automated tools to compare simulation results, and establish standards for kinematics, kinetics and muscle analysis (31), and not relying on users’ skills and experience.

6) OpenSim is a tool to study biomechanical variables that may have the potential to improve jump-landing tasks, such as muscle analysis. It is plausible that dancers’ technique may have a positive impact in jump-landing energy dissipation through the lower extremity. Not only joint angles, excursions, and moments, but also muscle activation and forces play an important part in contributing to the level of energy dissipation (25). Additional data of muscle analysis can be valuable to determine muscle activation patterns of the LE in jump-landing performance. Nonetheless, estimation of muscle forces and joint forces using musculoskeletal modeling involves a number of assumptions (94). Many modeling techniques to estimate muscle forces have been used, making challenging the selection of which one to be chosen.

8.4 Practical Implications

The implementation of training strategies in injury prevention programs has the principal aim to decrease musculoskeletal injuries while augmenting performance (60). Inappropriate control during landing tasks might lead to injuries (25). Athletes should be advised of the importance of landing techniques and the significance of landing safely (101). Biomechanical research has suggested that the forces imposed on the lower extremity joints can be modulated by changing the kinematic patterns (e.g., more or less lower extremity joint flexion), influencing the capability of modifying, absorbing and controlling the high impact forces associated with landing tasks (22, 48). Hence, altering the landing technique may be an appropriate intervention for injury prevention (67).

Previous studies (65, 94, 102, 103) reported changes in landing technique, such as increased lower extremity joint flexion corresponding to a softer landing, after specific instructions. Also, implementation of prevention programs has decreased the reliance on the knee extensors muscles and improved use of the hip extensor muscles, changing movement behavior at both hip and knee (65). Thus, appropriate instructions can alter the landing technique in athletes, and potentially reduce the risk of injuries. Based on this PhD thesis, an understanding of the kinematics and kinetics of dancers’ pattern provided useful information that may be relevant in injury mechanism and techniques for improving jump-landing performance (96). Some of the observed dance-related features in dancers jump-landing may suggest a positive impact in energy dissipation throughout the lower extremity. The training approaches used during the development of dance technique could therefore potentially offer some understanding and application into prevention programs in athletic populations (24). Lastly, beginning adequate jump- and balance-specific trainings at early ages may counteract the potentially deleterious adaptations in landing biomechanics observed in athletes (13).

Based on the findings of this PhD research, together with previous studies, it is suggested that to develop and implement intervention and injury prevention programs the following should be taken into consideration: 1) higher use of the lower extremity joint excursions (e.g. knee and foot-ankle complex), since the position at initial contact has an important effect for the following movement, and can provide a larger range of motion for the lower extremity joints as well as more time to dissipate the GRF; 2) higher focus on the foot and ankle angles at IC (more plantarflexed); 3) increased awareness of the lower extremity alignment, to avoid detrimental mechanisms such as sickling foot or dynamic knee valgus; and 4) incorporation of different jump-landing directions and base of supports in landings. Basically, the focus must be given on learn WHY and HOW to improve the landing technique and concomitant shock absorption mechanism; meanwhile engaging athletes in the programs, explaining and demonstrating the importance of include a prevention injury program in their daily work.

CHAPTER IX

9. Conclusions

It is believed that scientific knowledge is power to drive changes for better and alternative solutions. Overall, from the conducted research throughout this PhD work, not only relevant and scientific data regarding jump-landings were found, as well as prime information focused on multi-segmented foot model, and computational modeling and simulation in dancers. Based on this PhD thesis findings, it is suggested that there are dance-related characteristics that potentially may improve the absorption mechanism during landings, to successfully target modifiable risk factors. The findings also contribute to the body of knowledge in dance science. Further ahead, computational modeling and simulation can be an exceptional tool to accurately be applied in dancers’ jump-landing performances. Injury risk factors still require a larger in-depth biomechanical understanding, so the development and implementation of injury risk programs can become more efficient. Hence, based on the provided recommendations for future work (chapter VIII – section 8.3), additional research must continue examining jump-landings. In the end, the PhD experience is a milestone that marks the beginning of the research journey.

CHAPTER X

10. References

1. Russell JA, McEwan IM, Koutedakis Y, Wyon M. Clinical anatomy and biomechanics of the ankle in dance. J Dance Med Sci. 2008;12(3):75-82.
2. Hamilton WG, Hamilton LH, Marshall P, Molnar M. A profile of the musculoskeletal characteristics of elite professional ballet dancers. The American Journal of Sports Medicine. 1992;20.
3. Jacobs CL, Cassidy JD, Côté P, Boyle E, Ramel E, Ammendolia C, et al. Musculoskeletal injury in Professional dancers: prevalence and associated factors: an international cross-sectional study. Clini J Sports Med. 2016:1-8.
4. Caine D, Bergeron G, Goodwin BJ, Thomas J, Caine CG, Steinfeld S, et al. A Survey of Injuries Affecting Pre-Professional Ballet Dancers. J Dance Med Sci. 2016;20(3):115-26.
5. Allen N, Nevill A, Brooks J, Koutedakis Y, Wyon M. Ballet injuries: injury incidence and severity over 1 year. J Orthop Sports Phys Ther. 2012;42(9):781-90.
6. Ekegren CL, Quested R, Brodrick A. Injuries in pre-professional ballet dancers: Incidence, characteristics and consequences. J Sci Med Sport. 2014;17(3):271-5.
7. Hootman JM, Dick R, Agel J. Epidemiology of Collegiate Injuries for 15 Sports: Summary and Recommendations for Injury Prevention Initiatives. Journal of Athletic Training. 2007;42(2):311-9.
8. Agel J, Olson DE, Dick R, Arendt EA, Marshall SW, Sikka RS. Descriptive epidemiology of collegiate women's basketball injuries: NCAA Injury Surveillance System, 1988-1989 through 2003-2004. Journal of Athletic Training. 2007;42(2):202-10.
9. Agel J, Rockwood T, Klossner D. Collegiate ACL Injury Rates Across 15 Sports-National Collegiate Athletic Association Injury Surveillance System Data Update (2004-2005 Through 2012-2013. Clin J Sport Med. 2016;26(6):518-23.
10. Yang J, Tibbetts AS, Covassin T, Cheng G, Nayar S, Heiden E. Epidemiology of Overuse and Acute Injuries Among Competitive Collegiate Athletes. Journal of Athletic Training. 2012;47(2):198-204.
11. Liederbach M, Dilgen FE, Rose DJ. Incidence of anterior cruciate ligament injuries among elite ballet and modern dancers: a 5-year prospective study. Am J Sports Med. 2008;36(9):1779-88.
12. Gans A. The relationship of heel contact in ascent and descent from Jumps to the incidence of shin splints in ballet dancers. Physical Therapy. 1985;65(8):1192-6.
13. Orishimo KF, Kremenic IJ, Pappas E, Hagins M, Liederbach M. Comparison of landing biomechanics between male and female professional dancers. Am J Sports Med. 2009;37(11):2187-93.
14. Pollard CD, Sigward SM, Powers CM. Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments. Clin Biomech (Bristol, Avon). 2010;25(2):142-6.
15. Orishimo KF, Liederbach M, Kremenic IJ, Hagins M, Pappas E. Comparison of landing biomechanics between male and female dancers and athletes, part 1: Influence of sex on risk of anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1082-8.
16. Liederbach M, Kremenic IJ, Orishimo KF, Pappas E, Hagins M. Comparison of landing biomechanics between male and female dancers and athletes, part 2: Influence

of fatigue and implications for anterior cruciate ligament injury. Am J Sports Med. 2014;42(5):1089-95.
17. Blackburn JT, Padua DA. Influence of trunk flexion on hip and knee joint kinematics during a controlled drop landing. Clin Biomech (Bristol, Avon). 2008;23(3):313-9.
18. Blackburn JT, Padua DA. Sagittal-Plane trunk position, landing forces, and quadriceps electromyographic activity. Journal of Athletic Training. 2009;44(2):174-9.
19. Decker MJ, Torry MR, Wyland DJ, Sterett WI, Richard Steadman J. Gender differences in lower extremity kinematics, kinetics and energy absorption during landing. Clinical Biomechanics. 2003;18(7):662-9.
20. Schmitz RJ, Kulas AS, Perrin DH, Riemann BL, Shultz SJ. Sex differences in lower extremity biomechanics during single leg landings. Clin Biomech (Bristol, Avon). 2007;22(6):681-8.
21. Lee J, Song Y, Shin CS. Effect of the sagittal ankle angle at initial contact on energy dissipation in the lower extremity joints during a single-leg landing. Gait Posture. 2018;62:99-104.
22. De vita P, Skelly WA. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. 1992;24(1):108-15.
23. Agel J, Schisel J. Practice injury rates in collegiate sports. Clin J Sport Med. 2013;23(1):33-8.
24. Ward RE, Fong Yan A, Orishimo KF, Kremenic IJ, Hagins M, Liederbach M, et al. Comparison of lower limb stiffness between male and female dancers and athletes during drop jump landings. Scand J Med Sci Sports. 2018.
25. Yeow CH, Lee PV, Goh JC. An investigation of lower extremity energy dissipation strategies during single-leg and double-leg landing based on sagittal and frontal plane biomechanics. Hum Mov Sci. 2011;30(3):624-35.
26. Taylor JB, Ford KR, Schmitz RJ, Ross SE, Ackerman TA, Shultz SJ. Biomechanical differences in female basketball and soccer players during multi-directional jump landings. J Strength Cond Res. 2017;31(11):3034-45.
27. Taylor JB, Ford KR, Nguyen AD, Shultz SJ. Biomechanical Comparison of Single- and Double-Leg Jump Landings in the Sagittal and Frontal Plane. Orthop J Sports Med. 2016;4(6):1-9.
28. Donatelli R. Normal Biomechanics of the Foot and Ankle. J Orthop Sports Phys Ther. 1985;7(3):91-5.
29. Lin YC, Dorn TW, Schache AG, Pandy MG. Comparison of different methods for estimating muscle forces in human movement. Proc Inst Mech Eng H. 2012;226(2):103-12.
30. Lund ME, de Zee M, Andersen MS, Rasmussen J. On validation of multibody musculoskeletal models. Proc Inst Mech Eng H. 2012;226(2):82-94.
31. Hicks JL, Uchida TK, Seth A, Rajagopal A, Delp SL. Is My Model Good Enough? Best Practices for Verification and Validation of Musculoskeletal Models and Simulations of Movement. Journal of Biomechanical Engineering. 2015;137:1-24.
32. Seth A, Hicks JL, Uchida TK, Habib A, Dembia CL, Dunne JJ, et al. OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. PLoS Comput Biol. 2018;14(7):e1006223.
33. Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, et al. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng. 2007;54(11):1940-50.
34. Lund ME, Andersen MS, de Zee M, Rasmussen J. Scaling of musculoskeletal models from static and dynamic trials. International Biomechanics. 2015;2(1):1-11.
35. Trinler U, Baker R. Estimated landmark calibration of biomechanical models for inverse kinematics. Med Eng Phys. 2018;51:79-83.

36. Myers CA, Laz PJ, Shelburne KB, Davidson BS. A probabilistic approach to quantify the impact of uncertainty propagation in musculoskeletal simulations. Ann Biomed Eng. 2015;43(5):1098-111.
37. Ahonen J. Biomechanics of the Foot in Dance A Literature Review. J Dance Med Sci. 2008;12(3).
38. Kadel NJ. Foot and ankle injuries in dance. Phys Med Rehabil Clin N Am. 2006;17(4):813-26.
39. Sinsurin K, Vachalathiti R, Jalayondeja W, Limroongreungrat W. Different sagittal angles and moments of lower extremity joints during single-leg jump landing among various directions in basketball and voleyball athletes. J Phys Ther Sci. 2013;25:1109-13.
40. Sinsurin K, Vachalathiti R, Jalayondeja W, Limroongreungrat W. Altered Peak Knee Valgus during jump-landing among various directions in basketball and voleyball athletes. Asian Journal of Sports Medicine. 2013;4(3):195-200.
41. Self BP, Paine D. Ankle biomechanics during four landing techniques. Medicine & Science in Sports & Exercise. 2001;33(8):1338-44.
42. Hewett TE, Myer GD, Ford KR, Heidt RS, Jr., Colosimo AJ, McLean SG, et al. Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. Am J Sports Med. 2005;33(4):492-501.
43. Hewett TE, Myer GD, Ford KR. Anterior cruciate ligament injuries in female athletes: Part 1, mechanisms and risk factors. Am J Sports Med. 2006;34(2):299-311.
44. Tillman MD, Hass CJ, Brunt D, Bennett GR. Jumping and landing techniques in elite women's volleyball. J Sports Sci Med 2004;3:30-6.
45. Edwards S, Steele JR, McGhee DE. Does a drop landing represent a whole skill landing and is this moderated by fatigue? Scand J Med Sci Sports. 2010;20(3):516-23.
46. Shimokochi Y, Ambegaonkar JP, Meyer EG, Lee SY, Shultz SJ. Changing sagittal plane body position during single-leg landings influences the risk of non-contact anterior cruciate ligament injury. Knee Surg Sports Traumatol Arthrosc. 2013;21(4):888-97.
47. Kristianslund E, Krosshaug T. Comparison of drop jumps and sport-specific sidestep cutting: implications for anterior cruciate ligament injury risk screening. Am J Sports Med. 2013;41(3):684-8.
48. Zhang SN, Bates BT, Dufek JS. Contributions of lower extremity joints to energy dissipation during landings. Medicine & Science in Sports & Exercise. 2000;32(4):812-9.
49. LaBella CR, Huxford MR, Grissom J, Kim K, Peng J, Christoffel KK. Effect of Neuromuscular Warm-up on Injuries in Female Soccer and Basketball Athletes in Urban Public High Schools - Cluster Randomized Controlled Trial. Arch Pediatr Adolesc Med. 2011;165(11):1033-40.
50. Walden M, Atroshi I, Magnusson H, Wagner P, Hagglund M. Prevention of acute knee injuries in adolescent female football players: cluster randomised controlled trial. BMJ. 2012;344:e3042.
51. Fong C, Blackburn JT, Norcross MF, McGrath M, Padua DA. Ankle-dorsiflexion range of motion and landing biomechanics. J Athl Train. 2011;46(1):5-10.
52. McNitt-Gray JL. Kinematics and impulse chracteristics of drop landings from three heights. International Journal of Sport Biomechanics. 1991;7:201-24.
53. Cortes N, Onate J, Abrantes J, Gagen L, Dowling E, Van Lunen B. Effects of gender and foot-landing techniques on lower extremity kinematics during drop-jump landings. Journal of Applied Biomechanics. 2007;23:289-99.
54. Bahr R, Holme I. Risk factors for sports injuries—a methodological approach. Br J Sports Med. 2003;37:384-92.
55. Wilder RP, Sethi S. Overuse injuries: tendinopathies, stress fractures, compartment syndrome, and shin splints. Clinics in Sports Medicine. 2004;23(1):55-81.

56. Wiesler ER, Hunter DM, Martin DF, Curl WW, Hoen H. Flexibility and Injury Patterns in Dancers. Am J Sports Med. 1996;24(6):754-7.
57. Chockley C. Ground reaction force comparison between jumps landing on the full foot and jumps landing en Pointe in Ballet dancers. Journal of Dance Medicine & Science. 2008;12(1):5-8.
58. Murphy DF, Connolly DAJ, Beynnon BD. Risk factors for lower extremity injury: a review of the literature. Br J Sports Med. 2002;37:13-29.
59. Bonci CM. Assessment and Evaluation of Predisposing Factors to Anterior Cruciate Ligament Injury. Journal of Athletic Training. 1999;34(2):155-64.
60. Hewett TE, Bates NA. Preventive Biomechanics: A Paradigm Shift With a Translational Approach to Injury Prevention. Am J Sports Med. 2017;45(11):2654-64.
61. Aerts I, Cumps E, Verhagen E, Verschueren J, Meeusen R. A systematic review of different jump-landings. J Sports Med Phys Fitness. 2013;53:509-19.
62. Mason-Mackay AR, Whatman C, Reid D. The effect of reduced ankle dorsiflexion on lower extremity mechanics during landing: A systematic review. J Sci Med Sport. 2017;20(5):451-8.
63. Russell JA. Preventing dance injuries: current perspectives. Open Access J Sports Med. 2013;4:199-210.
64. Sinsurin K, Vachalathiti R, Jalayondeja W, Limroongreungrat W. Knee Muscular Control During Jump Landing in Multidirections. Asian J Sports Med. 2016;7(2):e31248.
65. Pollard CD, Sigward SM, Powers CM. ACL Injury Prevention Training Results in Modification of Hip and Knee Mechanics During a Drop-Landing Task. Orthop J Sports Med. 2017;5(9):1-7.
66. Jarvis DN, Kulig K. Pointing the foot without sickling: an examination of an ankle movement during jumping. Medical Problems of Performing Artistis. 2015.
67. Schot PK, Dufek JS. Landing performance part I: kinematic, kinetic, and neuromuscular aspects. Medicine, Exercise, Nutrition and Health. 1993;2:69-83.
68. Kovacs I, Tihanyi J, Devita P, Racz L, Barrier J, Hortobagyi T. Foot placement modifies kinematics and kinetics during drop jumping. Medicine & Science in Sports & Exercise. 1999;31(5):708-16.
69. Wyon MA, Twitchett E, Angioi M, Clarke F, Metsios G, Koutedakis Y. Time motion and video analysis of classical ballet and contemporary dance performance. Int J Sports Med. 2011;32(11):851-5.
70. Bowerman EA, Whatman C, Harris N, Bradshaw E. A review of the risk factors for lower extremity overuse injuries in young elite female ballet dancers. J Dance Med Sci. 2015;19(2):51-6.
71. Kulig K, Loudon JK, Popovich JM, Jr., Pollard CD, Winder BR. Dancers with achilles tendinopathy demonstrate altered lower extremity takeoff kinematics. J Orthop Sports Phys Ther. 2011;41(8):606-13.
72. Hoch MC, Farwell KE, Gaven SL, Weinhandl JT. Weight-Bearing Dorsiflexion Range of Motion and Landing Biomechanics in Individuals With Chronic Ankle Instability. J Athl Train. 2015;50(8):833-9.
73. Wikstrom EA, Tillman MD, Schenker SM, Borsa PA. Jump-landing direction influences dynamic postural stability scores. J Sci Med Sport. 2008;11(2):106-11.
74. Brockett CL, Chapman GJ. Biomechanics of the ankle. Orthopaedics and Trauma. 2016;30(3):232-8.
75. Okita N, Meyers SA, Challis JH, Sharkey NA. An objective evaluation of a segmented foot model. Gait Posture. 2009;30(1):27-34.
76. Towers JD, Deible CT, Golla SK. Foot and Ankle Biomechanics. Seminars in Musculoskeletal Radiology. 2003;7(1):67-74.
77. Stebbins J, Harrington M, Thompson N, Zavatsky A, Theologis T. Repeatability of a model for measuring multi-segment foot kinematics in children. Gait Posture. 2006;23(4):401-10.

78. Wright CJ, Arnold BL, Coffey TG, Pidcoe PE. Repeatability of the modified Oxford foot model during gait in healthy adults. Gait Posture. 2011;33(1):108-12.
79. Saltzman CL, Nawoczenski DA. Complexities of Foot Architecture as a Base of support. J Orthop Sports Phys Ther. 1995;21(6):354-60.
80. De Ridder R, Willems T, Vanrenterghem J, Robinson MA, Palmans T, Roosen P. Multi-segment foot landing kinematics in subjects with chronic ankle instability. Clin Biomech (Bristol, Avon). 2015;30(6):585-92.
81. Pothrat C, Authier G, Viehweger E, Berton E, Rao G. One- and multi-segment foot models lead to opposite results on ankle joint kinematics during gait: Implications for clinical assessment. Clin Biomech (Bristol, Avon). 2015;30(5):493-9.
82. Nester CJ, Liu AM, Ward E, Howard D, Cocheba J, Derrick T. Error in the description of foot kinematics due to violation of rigid body assumptions. J Biomech. 2010;43(4):666-72.
83. Russell JA, Shave RM, Kruse DW, Koutedakis Y, Wyon MA. Ankle and foot contributions to extreme plantar- and dorsiflexion in female ballet dancers. Foot Ankle Int. 2011;32(2):183-8.
84. Besier TF, Lloyd DG, Ackland TR. Muscle activation strategies at the knee during running and cutting maneuvers. Med Sci Sports Exerc. 2003;35(1):119-27.
85. Lundberg A. Kinematics of the ankle and foot: In vivo roentgen stereophotogrammetry. Acta Orthopaedica Scandinavica. 2009;60(sup233):1-26.
86. Carson MC, Harrington ME, Thompson N, O'Connor JJ, Theologis TN. Kinematic analysis of a multi-segment foot model for research and clinical applications: a repeatability analysis. Journal of Biomechanics. 2001;34:1299-307.
87. Milner CE, Brindle RA. Reliability and minimal detectable difference in multisegment foot kinematics during shod walking and running. Gait Posture. 2016;43:192-7.
88. Lucareli PR, Contani LB, Lima B, Rabelo ND, Ferreira CL, Lima FP, et al. Repeatability of a 3D multi-segment foot model during anterior and lateral step down tests. Gait Posture. 2016;43:9-16.
89. Lu TW, O'Connor JJ. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. J Biomech. 1999;32:129-34.
90. Leardini A, Belvedere C, Nardini F, Sancisi N, Conconi M, Parenti-Castelli V. Kinematic models of lower limb joints for musculo-skeletal modelling and optimization in gait analysis. J Biomech. 2017;62:77-86.
91. Valente G, Pitto L, Testi D, Seth A, Delp SL, Stagni R, et al. Are subject-specific musculoskeletal models robust to the uncertainties in parameter identification? PLoS One. 2014;9(11):e112625.
92. Krasnow D, Wilmerding V, Stecky S, Wyon M, Koutedakis Y. Biomechanical research in dance: a literature review. Medical Problems of Performing Artists. 2011.
93. Harwood A, Campbell A, Hendry D, Ng L, Wild CY. Differences in lower limb biomechanics between ballet dancers and non-dancers during functional landing tasks. Phys Ther Sport. 2018;32:180-6.
94. Laughlin WA, Weinhandl JT, Kernozek TW, Cobb SC, Keenan KG, O'Connor KM. The effects of single-leg landing technique on ACL loading. J Biomech. 2011;44(10):1845-51.
95. Mokhtarzadeh H, Yeow CH, Hong Goh JC, Oetomo D, Malekipour F, Lee PV. Contributions of the soleus and gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. J Biomech. 2013;46(11):1913-20.
96. Fong Yan A, Hiller C, Sinclair PJ, Smith RM. Kinematic analysis of sautes in barefoot and shod conditions. J Dance Med Sci. 2014;18(4):149-58.
97. Moniz-Pereira V, Cabral S, Carnide F, Veloso AP. Sensitivity of joint kinematics and kinetics to different pose estimation algorithms and joint constraints in the elderly. J Appl Biomech. 2014;30(3):446-60.

98. Cereatti A, Camomilla V, Vannozzi G, Cappozzo A. Propagation of the hip joint centre location error to the estimate of femur vs pelvis orientation using a constrained or an unconstrained approach. J Biomech. 2007;40(6):1228-34.
99. Schellenberg F, Taylor WR, Jonkers I, Lorenzetti S. Robustness of kinematic weighting and scaling concepts for musculoskeletal simulation. Comput Methods Biomech Biomed Engin. 2017;20(7):720-9.
100. Lees A, Vanrenterghem J, De Clercq D. Understanding how an arm swing enhances performance in the vertical jump. J Biomech. 2004;37(12):1929-40.
101. Briner WW, Kacmar L. Common injuries in volleyball. Mechanisms of injury, prevention and rehabilitation. Sports Med. 1997;24(1):65-71.
102. McNair PJ, Prapavessis H, Callender K. Decreasing landing forces- effect of instruction. Br J Sports Med. 2000;34:293-6.
103. Milner CE, Fairbrother JT, Srivatsan A, Zhang S. Simple verbal instruction improves knee biomechanics during landing in female athletes. Knee. 2012;19(4):399-403.