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Faculdade de Motricidade Humana

Filipa Oliveira da Silva João

Evaluation of mechanical load in the musculoskeletal system – development of experimental and modeling methodologies for the study of the effect of exercise in human models

Dissertação apresentada com vista à obtenção do grau de Doutor no ramo de Motricidade Humana, na especialidade de Biomecânica

Orientador: Doutor António Prieto Veloso

Presidente do Júri

Reitor da Universidade de Lisboa

Vogais

Doutor João Paulo Vilas Boas Soares Campos, Professor Catedrático da Faculdade de Desporto da Universidade do Porto

Doutor António Prieto Veloso, Professor Catedrático da Faculdade de Motricidade Humana da Universidade de Lisboa

Doutor Mário António Basto Forjaz Secca, Professor Associado da Faculdade de Ciências e Tecnologia da Universidade Nova de Lisboa

Doutor Pedro Victor Mil-Homens Ferreira Santos, Professor Associado da Faculdade de Motricidade Humana da Universidade de Lisboa

Doutor Paulo Rui Alves Fernandes, Professor Auxiliar com Agregação do Instituto Superior Técnico da Universidade de Lisboa

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Chapter 1

1 General Introduction

1.1 Introduction

This dissertation has a biomechanical basis where the main focus is related with sports movements, in particular jumping tasks characterized by explosive actions. To increase the knowledge and insight about the mechanisms that support this mechanical behavior, we explored different techniques and methods, applying them to a specific jumping exercise. The hopping model was chosen due to its wide use in musculoskeletal studies. From the motor control point of view, being an exclusive vertical jumping task, the structure of control is simpler and the optimization criterion is exclusively dependent on the vertical displacement of the centre of mass of the body. In addition, being unilateral, it is easy to control the utilization of the two legs separately, enabling the identification of possible compensatory mechanisms to control a fatiguing protocol. Specifically, to study the effect of a repetitive jumping task, the unilateral hopping has a high mechanical demand, thus the number of cycles needed to achieve a fatigue state is smaller, and, as consequence, less time period for testing. Moreover, being unilateral, there is only one force plate needed to collect the ground reaction force. Additionally, it allows us to study the propulsive capability of a unilateral jump, which has strong parallel to sports skills like running, volleyball and handball spike, athletics training drills, basketball shooting drills, etc.

1.1.1 Dissertation objectives

The first purpose of this dissertation was to investigate how different methodologies can provide us important information regarding the effect of mechanical loading in the lower limb musculoskeletal system. To investigate how those methodologies could be used and combined in a multidisciplinary approach, to better explain the mechanical behaviour of the lower limb musculoskeletal system, three complementary objectives were defined:

1) To investigate the effects of increasing fatigue on lower extremity biomechanical parameters using a traditional biomechanical analysis.

2) To use an induced acceleration approach to assess the contribution of the lower limb joints in general and individual muscles in particular, to accelerate the centre of mass of the body.

3) To use specific imaging techniques to assess the muscle architectural parameters that are more relevant to the production of force.

1.1.2 Dissertation overview

The present dissertation embraces a compilation of 5 articles published or submitted for publication in peer-review journals with ISI Impact Factor. In this context, each chapter is presented as an individual article following the format requested by the journal of submission/publication in respect to its sections. The overall purpose of this compilation was to investigate how different methodologies can provide us important information regarding the effect of mechanical loading in the lower limb musculoskeletal system, and how those methodologies can be combined in a multidisciplinary approach.

Accordingly, Chapter 3 contributes theoretically to the use of the stretch-shortening cycle model as a fundamental tool to study the effect of exercise, and particularly, the effect of fatigue achieved during cyclic movement, in the human lower limb mechanics. It comprises a manuscript (“**Neuromechanical changes in cyclic hopping exercise**”) aimed to present an overview of the literature on how the hopping model has been widely used to study the muscle-tendon behavior. It presents a full characterization of the kinematics, kinetics and electromyography (EMG) of the hopping exercise performed until fatigue.

The study presented in Chapter 4 (“**Quantifying the effect of plyometric hopping exercises on the musculoskeletal system: contributions of the lower limb joint moments of force to ground reaction forces in hopping exercise**”) aimed to estimate the ability of the lower limb joint moments of force to transfer mechanical energy through all the leg segments during the hopping task, performed until exhaustion. It was used an induced acceleration approach, which is a method based on the fact that the moments produced by muscle forces around a joint will accelerate all joints of the body and not just the joints they cross. Having this in mind, we analyzed how this contribution changes due to specific fatigue. An interesting question occurred afterwards, regarding the type of model used in this study: what if we changed the number of degrees of freedom of the model, what would happen to the contribution of each joint to propel the body upwards during the hopping? This leads us to Chapter 5 (“**Synergistic interaction between ankle and knee during hopping revealed through induced acceleration analysis**”), a manuscript where we emphasize the importance of the synergistic behavior of our body. In this case, we modeled the foot-floor interaction in two different ways: using a free or a fixed foot model. The intersegment coordination and synergistic behavior has tremendous consequences for the joint moments of force and, therefore, for the whole limb kinematics and kinetics. Therefore, for every question, we should look at the data and have a rationale based in the complexity of the task in order to choose the best set of model constraints.

In Chapter 6, ("**Muscle tendon behavior in single-leg cyclic hopping**") we wanted to gain more insight regarding the contribution of the muscle-skeletal system to accelerate the body not just at a net joint level, but specifically at an individual muscle level. In this case it was used a muscle induced acceleration analysis approach, with a model of the lower limb and torso with ninety two actuators, which contributed differently to the acceleration of the center of mass during the hopping.

Finally, Chapter 7 ("**Assessment of lower leg muscles architecture using diffusion tensor MRI: changes with cyclic stretch-shortening exercise**") appears in the sequence of this dissertation to answer some of the questions and limitations raised in Chapter 6, such as the subject-specific muscle characteristics that have more influence in the production of force: anatomical cross-section area, fiber length and pennation angle. In general, the muscle-skeletal models used in simulation software parameterize the muscles using information from cadaveric databases, therefore neglecting the individual muscle-tendon architectural and mechanical characteristics. Through the use of imaging techniques, namely magnetic resonance imaging (MRI), we quantified these parameters for the leg muscles in vivo, before and after cyclic hopping exercise until exhaustion.

1.2 Background

The knowledge of the mechanical properties of human muscle-tendon structure is extremely important in several ways:

1) to understand how the muscle–tendon units work and how they interact with each other, namely the energy changes between the contractile element and the series elastic element during the stretch shortening cycle (Anderson & Pandy, 1993; Ettema, 1996);
2) to obtain relevant data regarding input parameters for simulation models of the human motor system (Bobbert, 2001), and specifically how musculoskeletal geometry and musculo-tendon parameters affect muscle force and its moment about the joints (Delp et al., 1990);
3) to analyze the adaptation phenomena to different conditions like exercise type, fatigue, aging or injury, in order to better identify the specific factors that modify the performance capability (Blazevich, 2006; Bobbert, van der Krogt, van Doorn, & de Ruiter, 2011).

In sports movements, such as jumping and running, the ability to produce explosive actions is particularly important and there's an increased mechanical demand on the joints and tissues of the lower extremity. In the specific case of plyometric training, this mechanical demand produces not just neuromuscular developments (Komi, 2000) but also helps in the development of the technique and improvement of the control during impact (Markovic & Mikulic, 2010). This, in turn, will help to decrease the risk of injury in the lower limb (Mandelbaum et al., 2005) and may also induce structural adaptations on the bone and soft tissues (Kubo et al., 2007b).

1.2.1 Stretch-shortening cycle overview

Jumping activities are frequently chosen to study the mechanical properties of the muscle-tendon complex at different levels (from in vivo mechanical properties analysis to modeling and simulation) (Arampatzis, Schade, Walsh, & Bruggemann, 2001; Delp et al., 2007; Delp, et al., 1990; Fukashiro, Hay, & Nagano, 2006; Nagano, Komura, Yoshioka, & Fukashiro, 2005). Due to the easiness of performance and the relative low complexity of the movement, the hopping model is a widely used model to study muscle skeletal behavior. It can be performed with both legs or unilaterally. The latter case is the one discussed throughout this dissertation. One advantage of this type of hopping is the fact that there is only one force plate needed to collect the ground reaction force. Moreover, when thinking in protocols where the goal is to achieve a fatigue state, the unilateral hopping is profitable since there is less time needed for testing. The study of the propulsive capability of a unilateral hopping has also a strong parallel to sports skills like running, volleyball and handball spike, athletics training drills or basketball shooting drills. Hopping is commonly used in training routines to enhance athletic performance in terms of power, seeking the explosive reaction of the athlete through rapid eccentric contractions followed by powerful concentric contractions. Furthermore, being a vertical movement, the analysis is simplified because the control structure is simple and the vertical displacement of the centre of mass is the only variable needed for the optimization control.

The stretch-shortening cycle (SSC) is the mechanism of contraction behind plyometrics. Its efficiency requires a pre-activation of the muscles preceding the eccentric phase, a short and fast eccentric phase and a short time period between eccentric and concentric actions. The SSC has the purpose to improve the performance during the final phase (concentric action), when compared to the isolated concentric action (Komi, 2000). Most of the scientific evidence seems to indicate that the force potentiation of the SSC could be explained based on the storage of elastic energy during the stretching phase. In SSC movements, the elastic behavior of the muscle-tendon complex plays a key role and conflicting findings regarding muscle-tendon stiffness have been reported. Some authors suggest that a stiff muscle-tendon complex is optimal for SSC activities since it allows a rapid and efficient transmission of muscle force to the skeleton and higher rates of force development (Aura & Komi, 1986; Komi, 1986). Other studies reveal a negative correlation between muscle-tendon stiffness and the increase of performance in the concentric phase of the SSC (Kubo, Kawakami, Fukunaga, 1999; Kubo et al., 2007a; Stafilidis & Arampatzis, 2007). In this case, it is suggested that a more compliant muscle-tendon unit can store and release more elastic energy, which in turn could improve SSC performance by permitting the muscle fibers to operate at a more optimal length over the first part of their shortening range (Markovic & Mikulic, 2010).

The central and peripherical components of the neural control are crucial to the potentiation of the SSC. The muscle activation in the period before the contact with the ground (pre-activation) and during the reflex activation phases (in the late eccentric phase and early concentric phase) are extremely important. When dealing with repetitive SSC exercise, the literature shows conflicting findings. Some authors report no changes in the EMG activation pattern during pre-activation and eccentric phases (Kubo et al., 2007b), others report an increase in activity during the maximal voluntary contraction (Kyrolainen et al., 2005), and there is limited evidence of changes in the stretch-reflex excitability. Possible aspects for the neural adaptation to SSC exercise may be related with changes in leg muscle activation strategies and inter-muscular coordination patterns during jumping, as well as changes in the stretch reflex excitability (Markovic & Mikulic, 2010).

1.2.2 Traditional biomechanical analysis of human movement

Research has been investigating this area through the study of the kinematics, kinetics, and neuromuscular behavior, collecting data while performing these types of actions. All together, this information can provide an indication of the net summation of all muscle activity for each joint. (Bezodis, Kerwin, & Salo, 2008; Fukashiro, et al., 2006; Hunter, Marshall, & McNair, 2004; Weinhandl, Smith, & Dugan, 2011). Through the capture of the movement of the body, it is possible to assess the joint kinematics and using, for instance, a global optimization method, we can model the body segments adding it joint constraints. Simultaneously collecting the ground reaction forces, it is possible to analyze not only the three components of the ground reaction force vector but also use it to calculate the moments of force on each joint, through inverse dynamics. Here we have the opportunity to combine the observed motion of the segments with the ability of the muscles that act on those joints to produce the force necessary to perform that movement. Although inverse dynamics are common and well accepted in biomechanical analyses, it is a limited approach, from the standpoint of giving quantitative information regarding individual muscle function, since a net joint moment can be the result of several combinations of muscle forces (Schache et al., 2011).

Adding the electromyographic record of the main superficial muscles that participate in the studied movement, it is possible to assess the electrical activity of the muscles and infer about their neuromuscular behavior. The characterization of the neuromuscular activity during SSC has been extensively reported (Gollhofer, Komi, Fujitsuka, & Miyashita, 1987; Gollhofer & Kyrolainen, 1991; Horita, Komi, Nicol, & Kyrolainen, 1996; Komi, 2000; Kuitunen, Avela, Kyrolainen, & Komi, 2004; Kuitunen, Kyrolainen, Avela, & Komi, 2007; Nicol, Komi, Horita, Kyrolainen, & Takala, 1996) and used as another source of information to add to the kinematics and kinetics analysis.

Finally, the kinematics, kinetics and EMG data can be combined into musculoskeletal models, allowing the estimation of muscle forces or joint reaction forces. However, most of the computational models used have been created based on anatomical data collected from cadaveric specimens. To estimate muscle volume or PCSA, data of muscle force-generating capacities derived from cadaveric studies are often used to scale models of the muscle-tendon unit (Anderson & Pandy, 1999; Bobbert, 2001; Delp, et al., 1990). The use of those databases allows a better understanding of general principles of muscle function, but the comparability to individual muscle performance remains unclear (Albracht, Arampatzis, & Baltzopoulos, 2008).

If on one hand we want to analyze the human movement taking into account the body not just as a number of articulated segments that describe a certain behavior, but as a whole which works together in a synergistic way; on the other hand, we want to surpass the limitations that come from the use of standardized musculoskeletal models to obtain subject-specific information.

1.2.3 Induced acceleration in biomechanical analysis

Synergies occur in nature as an organization of elements which have the role of maintaining the stability properties of a certain performance variable (Latash, Gorniak, & Zatsiorsky, 2008). Muscles can generate force and develop mechanisms to exchange energy among segments, whether they are performing isometric, concentric or eccentric actions (Zajac, 2002; Zajac, Neptune, & Kautz, 2002). The knowledge of the exact role of a muscle in a particular movement is an important goal of the majority of researchers in the contexts of biomechanics, rehabilitation, performing arts, sports, etc.

Muscles are not always capable to directly deliver the energy that they produce to the target segment because its force does not accelerate that segment, instead acts to accelerate other segments. This means that the muscle works in a synergistic way, producing energy so that other muscles can use it to accelerate or decelerate segments in order to reach the target segment (Zajac, 2002). Induced acceleration analysis (IAA) is a technique based in the dynamic coupling effect caused by the multi-articulated nature of the body (Zajac, 1993, 2002; Zajac & Gordon, 1989). Dynamic coupling means that when a muscle contracts it produces acceleration, not only in those segments that are spanned by that muscle but on all body segments of the chain, due to the intersegment forces. Thus, this technique allows the direct quantification of a joint moment contribution to the acceleration of each joint of the body and to the body center of mass.

Analyses of induced accelerations have been mainly used with normal gait (Correa, Crossley, Kim, & Pandy, 2010; Correa & Pandy, 2012; Dorn, Lin, & Pandy, 2012; Goldberg & Kepple, 2009; Liu, Anderson, Pandy, & Delp, 2006; Liu, Anderson, Schwartz, & Delp, 2008; Neptune, Kautz, & Zajac, 2001; Neptune & McGowan, 2011; Pandy & Andriacchi, 2010; Yi-Chung Lin, 2011), impaired gait (Arnold, Anderson, Pandy, & Delp, 2005; Siegel, Kepple, & Stanhope, 2006, 2007), running (Hamner, Seth, & Delp, 2010; Neptune & McGowan, 2011) and pedaling (Neptune, Kautz, & Zajac, 2000), in order to study the changes in muscle contributions to motion. However, most dynamic human movements in sports activities involve cyclic stretch-shortening actions with maximal mechanical power production, so this technique should be of benefit in analyzing the different contributions of lower limb muscle groups over a time period during cyclic exercise. This tool could also be useful to assess muscle groups adaptive compensations when muscle fatigue occurs. It is being clearly showed by several authors (Goldberg & Kepple, 2009; Kepple et al., 1997; Siegel, et al., 2007) that the effect of a muscle group action or a joint moment of force can accelerate body segments that are located far on the segmental chain through a coupling effect.

1.2.4 Imaging techniques in biomechanical analysis

Musculoskeletal models allow us to estimate, amongst other parameters, the force-generating capacity of muscles (that are difficult to measure experimentally). Recently, models of the musculoskeletal system comprise the tridimensional surface geometry of bones, the kinematics of the joints and a considerable number of actuators that represent the muscles and tendons to which were assigned mechanical characteristics, acquired from cadaveric databases. The combination of these models with dynamic simulation has been used to gain more insight into normal and pathological movement (Anderson & Pandy, 2003).

One limitation of musculoskeletal models is the input parameters which typically are based in cadaveric data obtained from different studies, usually with a few numbers of specimens (Friederich & Brand, 1990; Wickiewicz, Roy, Powell, & Edgerton, 1983). It is likely that data from cadaveric specimens do not accurately reflect absolute or relative sizes of muscles in young and healthy subjects (Tate, Williams, Barrance, & Buchanan, 2006). Another limitation is the simplification of some muscle architectural parameters, such as muscle fiber length (normally are averaged in cadaveric studies) or pennation angle (common models assume muscle fibers as one single structure with only one angle between the fibers direction and the aponeurosis). For this reason, the need to create accurate, individualized models of the musculoskeletal system is driving advances in imaging techniques, especially in MRI (Blemker, Asakawa, Gold, & Delp, 2007).

MRI is considered the most useful non-invasive imaging technique that allows the reconstruction of the whole muscle, enabling the quantification of detailed structure, function and metabolism from the muscle tissue. One of the MRI methods is the Diffusion tensor imaging (DTI). This imaging technique relies on the correspondence between cell geometry and the anisotropic nature of water diffusion in the muscle (Heemskerk, Sinha, Wilson, Ding, & Damon, 2010). The theoretical basis for DTI states that self-diffusion of water in tissue is restricted by membranes and other cellular constituents, resulting in an apparent diffusion coefficient, which is lower than the free diffusion coefficient and is orientation-dependent for elongated structures. Skeletal muscles are composed of fascicles that have an elongated, generally cylindrical shape. The three principal axes of the diffusion tensor can be calculated and give us information regarding the shape and the direction of the diffusion (Noseworthy, Davis, & Elzibak, 2010). Through a tractographic process, it is possible to quantify the number of fascicles, estimate fascicle length and also observe and quantify the different pennations in different areas of the muscle. Throughout this dissertation, it will be used the word “fiber” instead of fascicle, when we address the issues related with diffusion tensor imaging and tractography. The reason lies in the fact that is common to use this term in the journals from this research field, namely magnetic resonance imaging publications.

Another MRI test used is the signal obtained from Proton density (PD) weighted scans, which is directly proportional to the number of available spins (hydrogen nuclei in water). It consists in a spin echo or a gradient echo sequence, with short echo time and long repetition time. PD images are useful for determining the muscles boundaries, due to the elevated detail in the image and tissue differentiation capability, enabling the determination of muscle cross sectional area, and subsequently muscle volume using simple image processing software.

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Chapter 2

2 Methodological Considerations

The subsections of this chapter are a brief description of each one of the methods used to accomplish the purposes mentioned in the Introduction. Nevertheless, in each chapter (from three to seven), the methodology will be described more in detail, according to the specific objective of the chapter.

2.1 Motion Capture and 3D modeling reconstruction

This subsection refers to the methods used to obtain the data that serve as input in all of the thesis' chapters. To record the hopping movement, a tridimensional motion capture was collected in the Biomechanics Laboratory with an optoelectronic system of ten cameras Qualisys (Ocqus 300 model) operating at 200Hz. Twenty-four reflective markers and four marker clusters were placed in predefined anatomical protuberances (acromial, stern, anterior-superior and posterior-superior iliac spines, lateral and medial femoral condyles, lateral malleoli of the fibula, medial malleoli of the tibia, 1st and 5th metatarsal heads and one marker cluster for each thigh, shank and foot segments), as can be seen in Fig. 2-1A. These markers were used for the reconstruction of eight body segments (trunk, pelvis, right and left thighs, right and left shanks and right and left feet) in Visual 3D software (Visual 3D Basic RT, C-Motion, Inc., Germantown, MD) (Fig. 2-1B).

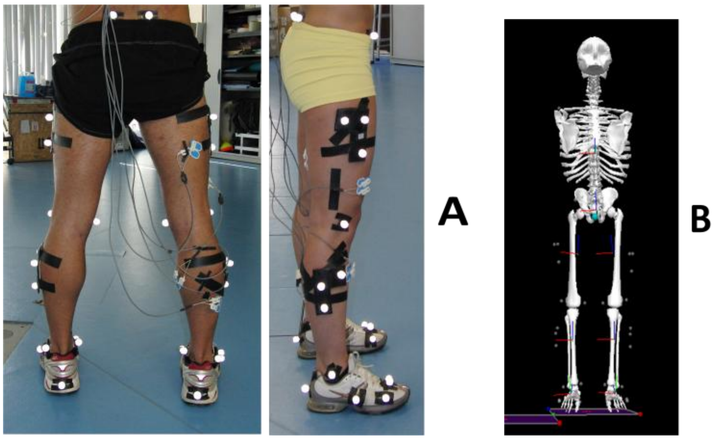

Figure 2-1: A) Segments' reflective markers setup and B) segment's model reconstruction in Visual 3D

Each body segment was reconstructed as a rigid body, with a length and an embedded coordinate system. Visual 3D assumes that each segment moves accordingly with the motion of the tracking markers that are attached to it. Regarding this, the system tracks the position and orientation of each segment, allowing for the calculation of all the kinematic variables. We used a global optimization method that is based on the search of an optimal pose of the multilink model for each data frame such that the overall differences between the measured and model-determined marker coordinates are minimized in a least square sense, throughout all the body segments. This method, also known as inverse kinematics (IK), considers measurement error distributions in the system and provides an error compensation mechanism between body segments, which can be regarded as a global optimization at the system level (Lu & O'Connor, 1999). Basically, it relies on the determination of a pre-defined kinematic model with predefined joint constraints (Andersen, Damsgaard, & Rasmussen, 2009; Duprey, Cheze, & Dumas, 2010). The joint constraints can be added between segments and restrict the relative motion between them. This is accomplished by creating one or more IK chains. The IK chain must start with a root segment, which in this study was the pelvis segment. A root segment has the global coordinate system as a parent segment and is connected to the laboratory with six degrees of freedom. Once the root segment is added, the remainder of the IK chain can be built. When building the IK model we need to specify the motion that is permitted between two segments. The hip joint was defined as a spherical joint with 3 degrees of freedom (flexion/extension, abduction/adduction and internal/external rotation); the knee was defined as a pin joint with one degree of freedom (flexion/extension) and the ankle was defined as a universal joint with 2 degrees of freedom (dorsi/plantaflexion and internal/external rotation).

For the kinetic data calculation, the mass properties are based on Dempster's (1955) anthropometric data and the moments of inertia properties are based on Hanavan's study (1964). Ground reaction force (GRF) was collected with a Kistler force plate (type: 9865B).

2.2 Electromyography

This subsection refers to the methodology used to obtain the neuromuscular data analyzed more in detail in chapters three and six. The electric activity from five muscles of the dominant lower limb was recorded: tibialis anterior (TA), gastrocnemius medialis (GM), soleus (SOL), vastus lateralis (VL) and biceps femoris (BF). The participant's skin was shaved, gentle abraded and cleaned with alcohol. Afterwards, the surface electrodes (Ambu Blue Sensor N-00-S/25) were placed with an inter-electrode distance of 20mm, in accordance with the SENIAM Project recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). TA: electrodes placed at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus. GM: electrodes placed on the most prominent bulge of the muscle in the direction of the leg. SOL: electrodes placed at 2/3 of the line between the medial condylis of the femur to the medial malleolus. VL: electrodes placed at 2/3 on the line from the anterior superior spina iliaca to the lateral side of the patella. BF: electrodes placed at 50% in the direction of the line between the ischial tuberosity and the lateral epicondyle of the tibia. A ground electrode was placed over the C7 vertebrae. The EMG data were transmitted using a telemetric system (Glonner Biotel 88) and collected at 1kHz. The EMG signals were processed through a 32bit A/D board synchronized with the force platform and the motion capture system. The processing of the EMG data is explained more in detail in the correspondent chapters.

2.3 Segmental and muscle induced acceleration analysis

In this subsection we intend to explain the induced acceleration analysis (IAA), performed using a model without muscles and another one with muscles. IAA is a technique based in the dynamic coupling effect caused by the multi-articulated nature of the body (Zajac, 1993, 2002; Zajac & Gordon, 1989). Dynamic coupling means that when a muscle contracts it produces acceleration, not only in those segments that are spanned by that muscle but on all body segments of the chain, due to the intersegment forces. Thus, this technique allows the direct quantification of a joint moment contribution to the acceleration of each joint of the body and to the acceleration of the body center of mass. The dynamic equations of motion can be expressed in the following form:

(Eq.2-1)

where s the joint accelerations matrix, is the inverse inertia matrix (where the segments inertial parameters and center of mass positions are taken into account), is the joint moments matrix, is the Coriolis terms matrix and is the Gravitational terms matrix. To isolate the contribution of one particular joint moment to the acceleration of all the joints of the model it is assumed that all the other joints have moments and stiffness equal to zero. Given equation 2-1, and terms are set to zero allowing us to obtain the accelerations produced by each one of the joint moments (Kepple, Siegel, & Stanhope, 1997). The magnitude of the computed accelerations is dependent, not only on the moment magnitude (τ), but also on the configuration of the body segments (θ):

(Eq.2-2)

In chapter 6, a model with muscle actuators was used. For this case, the IAA was used to compute the contributions of individual muscles to the acceleration of the body mass center (Hamner, Seth, & Delp, 2010; Zajac & Gordon, 1989). The equations of motion for multibody dynamic systems were solved.

(Eq.2-3)

where is a vector of inertial forces and moments; is a vector of centrifugal and Coriolis forces and moments; is a vector of gravitacional forces and moments; is a matrix of muscle moment arms; is a vector of muscle forces; and is a vector of external forces and moments applied to the body by the environment. And this equation, due to the properties of the mass matrix , can be solved for the accelerations that cause each of the forces as follows (Pandy & Andriacchi, 2010):

(Eq.2-4)

2.4 Magnetic Resonance Imaging techniques: diffusion tensor imaging (DTI) and proton density (PD)

DTI technique relies on the correspondence between cell geometry and the anisotropic nature of water diffusion in the muscle (Heemskerk, Sinha, Wilson, Ding, & Damon, 2010). The theoretical basis for DTI states that self-diffusion of water in tissue is restricted by membranes and other cellular constituents, resulting in an apparent diffusion coefficient, which is lower than the free diffusion coefficient and is orientation-dependent for elongated structures. The diffusion tensor can be mathematically described by a 3x3 matrix, and at least six independent directions for the diffusion gradients must be assessed to calculate the diffusion tensor. The three principal axes of the diffusion tensor can be calculated through a diagonalization process, where the eigenvalues and eigenvectors are determined. These parameters give us information regarding the shape and the direction of the diffusion, respectively (Noseworthy, Davis, & Elzibak, 2010). Other parameters, derived from these, can be quantified: the fractional anisotropy (FA) and the apparent diffusion coefficient (ADC). From the tracking of the fibers, it is possible to calculate the average fiber length and the pennation angle.

The signal obtained on Proton density weighted scans is directly proportional to the number of available spins (hydrogen nuclei in water). It consists in a spin echo or a gradient echo sequence, with short echo time and long repetition time. PD images are useful for determining the muscles boundaries, due to the elevated detail in the image and tissue differentiation capability, enabling the determination of muscle cross sectional area, and subsequently muscle volume using simple image processing software.

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Chapter 3

3 Neuromechanical changes in cyclic hopping exercise[1]

[1] Submitted as:

Filipa João & António Veloso. Neuromechanical changes in cyclic hopping exercise. Submitted to the European Journal of Sports Sciences (under review).

3.1 Abstract

The purpose of this study was to investigate the effects of increasing fatigue on lower extremity joint kinematics, kinetics and electromyographic parameters during a cyclic unilateral jumping exercise. Nine young active adults performed repetitive unilateral jumps with their dominant limb until complete exhaustion. With fatigue, the subjects landed with the knee more flexed and the ankle more dorsi flexed. The vertical ground reaction force peak decreased, as well as the ankle and knee peak joint moments of force. The joint power profiles show a decrease in both the ankle and knee absorption and propulsion peaks but no significant differences at the hip joint. These results are in agreement with the reduction in the electromyography bursts of the plantar flexors, which are expected for an exhaustive stretch-shortening cycle. At the same time, the biceps femoris muscle significantly increased its activation in the reflex-induced activation phase, which may indicate a reduction in the knee extensors load with the fatigue condition. No differences were found in joint stiffness, although the vertical stiffness increased. Repetitive stretch-shortening cycle exercise appears to induce a reduction in force and power production and using the unilateral hopping model it is possible to study the fatigue phenomena and observe the alterations in the kinematics, kinetics and neuromuscular pattern of the lower extremity.

3.2 Introduction

In sports movements, such as jumping, the ability to produce explosive actions is extremely important and there's an increased mechanical demand on the joints and tissues of the lower extremity. Markovic & Mikulic (2010) review on plyometric training, emphasizes that this type of exercise has been used to improve neuromuscular function (Komi, 2000; Markovic, Jukic, Milanovic, & Metikos, 2007), to improve biomechanical technique and neuromuscular control during high impact tasks (Myer, Ford, Palumbo, & Hewett, 2005), to reduce the risk of lower extremity injuries in team sports (Mandelbaum et al., 2005) and to induce bone, muscle and tendon adaptations (Kubo et al., 2007). The stretch-shortening cycle (SSC) is the mechanism of contraction behind plyometrics. Its efficiency requires a pre-activation of the muscles preceding the eccentric phase, a short and fast eccentric phase and a short time period between eccentric and concentric actions (Veloso, Pezarat-Correia, Armada, & Abrantes, 1998). A pre-stretch improves the force and work output that muscles can produce during the concentric phase, due to the combination of the following aspects: a) the time to develop force, enhancing the performance during the concentric phase; b) the storage and reutilization of elastic energy; c) interaction between the series elastic components and the contractile component and d) the contribution of reflexes (Markovic & Mikulic, 2010).

Most of the studies concerning SSC mechanics analyze kinematic and kinetic variables such as joint angles and velocities, muscle net moments, kinetic energy, joint powers, individual or net muscle activation and joint mechanical work (Bezodis, Kerwin, & Salo, 2008; Hunter, Marshall, & McNair, 2004; Johnson & Buckley, 2001; Novacheck, 1998; Slawinski et al., 2010). All together, these variables can provide an indication of the net summation of all muscle activity for each joint. SSC is also a naturally occurring action in the human movement therefore being relevant its analysis during the fatigue phenomena.

The hopping exercise is a widely used model to study muscle skeletal behavior. Being a unilateral movement, with high mechanical demand, the number of cycles needed to achieve a fatigue state is smaller, and, as consequence, less time period for testing. Moreover, there are fewer possibilities to use the other leg as a compensatory mechanism to control the fatiguing condition. Additionally, it allows us to study the propulsive capability of a unilateral jump, which has strong parallel to sports skills like running, volleyball and handball spike, athletics training drills, basketball shooting drills, etc.

Fatigue is a complex condition well reported in several studies but yet quite difficult to quantify. Citing Enoka & Duchateau (Enoka & Duchateau, 2008), muscle fatigue «can refer to a motor deficit, a perception or decline in mental function, it can describe the gradual decrease in the force capacity of muscle or the endpoint of a sustained activity, and it can be measured as a reduction in muscle force, a change in electromyographic activity or an exhaustion of contractile function». The effects of neuromuscular fatigue on lower extremity biomechanics may be a contributing factor in musculoskeletal injuries during prolonged activity, leading the body to acquire compensatory strategies to execute the movement. Several studies regarding fatiguing SSC actions (Gollhofer, Komi, Miyashita, & Aura, 1987; Horita, Komi, Nicol, & Kyrolainen, 1996) showed that there is a shift in muscle activation from the braking phase to the push-off phase with fatigue, which may be an indicator of the inability of the muscle-skeletal system to maintain the impact loads and the storage of elastic energy (Gollhofer, Komi, Fujitsuka, & Miyashita, 1987). And consequently, as mentioned by Kuitunen, Kyrolainen, Avela, & Komi (2007), there is a demand in more energy to convert from the metabolic sources, to maintain the power output during the push-off phase, which sequentially leads to a faster progression of the fatigue state. This means that we have changes in the neuromuscular parameters (reflected in the EMG activity) and in soft tissue mechanics (reflect in joint and vertical stiffness). It is difficult to assess which parameter is more relevant or presents a more significant contribution to the mechanical changes observed in the fatigue condition therefore we analyzed the behavior of both.

Both linear and angular kinematics, as well as ground reaction force and net joint moments of force assessment, enable us the establishment of an analogy between the hopping and mass-spring models. The concept of stiffness is now introduced, based on Hooke's law which states that the force required to deform an object is related to an elastic constant (spring) and the distance of deformation of the object (Butler, Crowell, & Davis, 2003). Regarding this assumption of the body segments as springs, it is possible to describe the ability of the body to resist displacement once ground reaction force or moments are applied (Serpell, Ball, Scarvell, & Smith, 2012). Vertical stiffness is the reference stiffness measure and is calculated through the quotient of maximum ground reaction force and centre of mass displacement (Butler, et al., 2003). Joint stiffness is the resistance to change in angular displacement with the application of a joint moment. Several studies have pointed out that both ankle and knee joints have important roles in the leg stiffness modulation (Arampatzis, Schade, Walsh, & Bruggemann, 2001; Farley & Morgenroth, 1999; Horita, et al., 1996; Kuitunen, Avela, Kyrolainen, Nicol, & Komi, 2002), sometimes with contradictory or unclear results, but overall research has related stiffness with sports performance, particularly for force transmission, which should be the most efficient for a certain sportive outcome.

The kinematic changes observed on a fatigued athlete are the result of muscle performance impairments that contribute to the athlete's inability to maintain the same mechanical output for a long period of time. Those changes may increase injury risk potential, especially in athletes with less experience or muscle imbalances (Gerlach et al., 2005). The design of an appropriate training program which improves the athlete's performance without compromising its musculoskeletal system is, consequently, important. In order to achieve this, we need to understand how the muscle-tendon unit changes its ability to perform work when the body is fatigued. Therefore, experimental protocols that induce fatigue need to be designed and the effect that fatigue has on the lower-limb musculoskeletal system needs to be documented.

The purpose of this study was to perform a complete characterization of the kinematics, kinetics and neuromuscular functioning of the lower limb during a unilateral jumping exercise (hopping) performed continuously until mechanical failure. To characterize this task during the entire hopping time, 10 hops separated in time were chosen from the total sequence (1 hop every 10% of the total number of hops) for analysis and comparison.

3.3 Methods

3.3.1 Participants

Nine healthy active students, male (n=4, years, 66.4 ± 7.4 kg, m) and female (n=5, years, 61.7 ±9.9 kg, m) participated in this study. All subjects were involved in competitive or recreational sports and reported no history of lower extremity injury within the previous 12 months. Informed consent was given by each subject prior to testing, and approval for the protocol was obtained from the Ethical Committee of the Faculdade de Motricidade Humana.

3.3.2 Exercise protocol

After a warm up, the subjects positioned with one foot on the force plate and kept their hands on their waist. The task consisted in a hopping sequence with the dominant lower limb over a force plate, until complete failure. To establish a control parameter for the jump height, three maximal effort squat jumps (SQJ) were performed before the task, to estimate jump height based on the vertical displacement of the body's center of mass. The criteria chosen for the hopping performance was the minimum height, which was set at 60% of the maximum height achieved in the SQJs (it was set 60% because unilateral hopping is mechanically more demanding thus the number of hops was not too high and the task was more controlled). The subjects performed both the SQJs and the hops receiving visual feedback about jump high, being informed with a red color if jumping high was insufficient. This contact mat was used as an instantaneous system for the subjects to see their jumping height. They were instructed to keep the hands on the waist to minimize arm motion because the head, arms and trunk were modeled as a single segment (Kepple, Siegel, & Stanhope, 1997). The hopping frequency was freely chosen by the subjects. One hop cycle began with the contact of the right foot on the ground, continued through the suspension phase and ended with the next right foot contact. The exercise was stopped when the subjects could no longer achieve the 60% height systematically, even though they were encouraged to continue the hopping until they could no longer take their foot off the ground.

3.3.3 Data acquisition

Motion capture was collected with 10 cameras (Qualisys Oqus-300 and Qualisys Track Manager, Qualisys AB, Gothenburg, Sweden ®) operating at 200Hz. 24 reflective markers and 4 marker clusters were placed in predefined anatomical protuberances (acromial, stern, anterior-superior and posterior-superior iliac spines, lateral and medial femoral condyles, lateral malleoli of the fibula, medial malleoli of the tibia, 1st and 5th metatarsal heads and one marker cluster for each thigh, shank and foot segments) and used for the reconstruction of eight body segments (trunk, pelvis, right and left thighs, right and left shanks and right and left feet) using Visual 3D software (Visual 3D Basic RT – Version 4.94.0, C-Motion, Inc ®). The moment of inertia properties were based on Hanavan's study (Hanavan, 1964) and the mass properties were based on Dempster's (1955) anthropometric study.

Ground reaction force (GRF) was collected with a Kistler force plate (type: 9865B). Kinematic and kinetic variables were processed with a lowpass 4th order Butterworth filter with a frequency cut of 6 Hz. The curves were plotted and normalized to the stance phase of the hop. For each dynamic trial, Visual 3D was used to calculate both kinematic and kinetic parameters. A global optimization process (inverse kinematics) was used to minimize the effect of soft tissue and measurement errors with the following joint constraints: the ankle was designed as a universal joint (plantar/dorsi flexion and inversion/eversion), the knee as a revolute joint (flexion/extension) and the hip as a spherical joint (flexion/extension, abduction/adduction and internal/external rotation). The net joint moments were calculated through inverse dynamics. Ankle, knee and hip powers were also calculated as the product of the instantaneous joint moment and joint angular velocity. Mechanical joint work, defined as the integral of the joint power curve, was calculated for the three joints during the contact phase of the hop. All kinetic parameters were normalized to the subject's body mass. Lower limb stiffness was calculated during the contact phase by dividing the GRF peak by the vertical displacement of the center of mass of the subject. Joint stiffness was calculated by dividing the correspondent joint moment of force peak by the angular displacement during the contact phase.

Electromyographic (EMG) signals from 5 muscles from the dominant lower leg were recorded: tibialis anterior (TA), gastrocnemius medialis (GM), soleus (SOL), vastus lateralis (VL) and biceps femoris (BF). The participant's skin was shaved, gentle abraded and cleaned with alcohol. Afterwards, the surface electrodes (Ambu Blue Sensor N-00-S/25) were placed with an interelectrode distance of 20mm, in accordance with the SENIAM Project recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). TA: electrodes placed at 1/3 on the line between the tip of the fibula and the tip of the medial malleolus. GM: electrodes placed on the most prominent bulge of the muscle in the direction of the leg. SOL: electrodes placed at 2/3 of the line between the medial condylis of the femur to the medial malleolus. VL: electrodes placed at 2/3 on the line from the anterior spina iliaca superior to the lateral side of the patella. BF: electrodes placed at 50% in the direction of the line between the ischial tuberosity and the lateral epicondyle of the tibia. A ground electrode was placed over the C7 vertebrae. The EMG data were transmitted using a telemetric system (Glonner Biotel 88) and collected at 1kHz (Biopac MP100). The EMG signals were processed through a 32bit A/D board synchronized with the force platform and motion capture system. The EMG signals were full wave rectified and integrated (IEMG) over different functional phases: pre-activation (100ms prior to contact), reflex induced activation phase (RIA) (from 30ms to 120ms after contact) and late EMG response phase (the latest 120ms of the contact phase). The RIA phase was defined in agreement with the short and medium latency stretch reflexes for VL (Horita, et al., 1996). The EMG signal was normalized to peak activity recorded during a maximal squat jump (SQJ).

To accommodate for inter subject variation in the total of hops completed (which averaged 107.7±51.5), we divided the complete exercise in ten equal intervals. It was analyzed 1 representative hop at each 10% interval. This hop was chosen looking at the mean value of the hop heights in that interval. The one which more closely matched the mean value of the height was selected. The 10 hops of each subject were then analyzed to determine the effect of repetition on the joint mechanics during the contact phase.

3.3.4 Statistical analysis

The variables analyzed in this study were: ankle, knee and hip joint positions at initial contact (IC), toe-off (TO) and joint range of motion (ROM) during the contact phase; peak vertical GRF, peak moments of force, maxima and minima from the joint power profiles, negative and positive joint works and EMG parameters such as pre-activation (PA), reflex-induced activation (RIA) and late EMG response (LER). The kinematic, kinetic and EMG parameters were compared using repeated-measures ANOVA to analyze the differences between the 10 representative hops. Statistical significance was set at p-value <0.05 and all statistical analyses were performed using SPSS (version 20, SPSS Science Inc., Chicago, IL, USA).

3.4 Results

The subjects performed an average of 107.7±51.5 hops during the entire sequence. The hop selected during the first interval of the hopping sequence will be referred as [0%-10%] and the last corresponds to [90%-100%]. Regarding the contact time, peak of vertical GRF, impulse of force and hop height, we verified significant decreases from the first to the last hop, except for the contact time. Concerning joint kinematics, it was analyzed the angle at initial contact (IC), toe-off (TO), minimum angle (min) and range of motion (ROM) for the three joints (Table 1) in the sagittal plane. As a result of fatigue, the subjects landed in the ground with the foot more dorsiflexed and at the toe off it was less plantarflexed. The ankle amplitude also decreased significantly. The knee became more flexed at the IC and TO, with a decrease in the total range of motion. The hip joint revealed a significant increase in flexion at the IC.

Table 3-1: Mean and standard deviation of the kinematic and kinetic parameters assessed in the first [0%-10%] and last [0%-100%] representative hops of the exercise. Significant differences are referred with an* and the correspondent p-value.

	[0%-10%]	[90%-100%]	p-value
Angular kinematics (deg)			
Ankle ROM	46.2±4.2	32.6±4.7*	0.000
Ankle IC	55.3±8.3	64.4±5.6*	0.004
Ankle TO	48.1±7.5	58.9±4.8*	0.002
Ankle min	95.6±6.9	91.3±4.8	
Knee ROM	44.3±9.8	35.1±7.6*	0.032
Knee IC	18.1±5.6	30.3±6.3*	0.000
Knee TO	6.3±3.9	12.8±5.6*	0.008
Knee min	6.3±3.9	12.8±5.5*	0.008
Hip ROM	21.2±8.4	21.9±9.3	
Hip IC	22.0±4.5	29.4±4.9*	0.001
Hip TO	14.5±5.3	17.3±6.5	
Hip min	14.2±5.3	16.7±5.7	
Peak GRF (N/kg)	26.7±3.5	24.7±3.0*	0.006
Contact Time (s)	0.376±0.048	0.374±0.059	
CM height (m)	1.03±0.04	0.95±0.06*	0.001
Impulse (N.s/kg)	5.82±0.65	5.14±0.51*	0.000
Peak Moments of force (Nm/kg)			
Ankle	2.7±0.6	2.2±0.5*	0.001
Knee	3.3±0.4	2.8±0.4*	0.000
Hip	1.9±0.4	2.1±0.4	
Peak joint Negative Powers (W/kg)			
Ankle	-8.7±2.3	-5.4±1.9*	0.001
Knee	-9.6±3.1	-6.7±1.8*	0.021
Hip	-2.5±1.2	-2.8±0.6	
Peak joint Positive Powers (W/kg)			
Ankle	8.8±1.5	4.7±1.0*	0.000
Knee	8.8±1.6	6.3±1.2*	0.019

Hip	2.7±1.3	2.8±1.0	
Joint negative work during contact phase (J/kg)			
Ankle	0.74±0.19	0.42±0.13*	0.002
Knee	0.73±0.31	0.45±0.20*	0.001
Hip	0.70±0.22	0.83±0.23*	0.000
Joint positive work during contact phase (J/kg)			
Ankle	0.84±0.13	0.47±0.09*	0.002
Knee	0.77±0.25	0.53±0.13*	0.015
Hip	0.77±0.15	0.92±0.17*	0.000
Vertical stiffness (KN/m)	9.76±0.87	11.36±0.89*	0.013
Joint stiffness (Nm/deg)			
Ankle	4.52±0.54	5.00±0.62	
Knee	7.20±1.25	10.86±2.83	
Hip	7.89±1.48	9.03±3.12	

The electromyographic pattern of the five muscles is presented in Figure 3-1, in accordance with the kinematics of the ankle and knee joints, as well as the GRF curves. The data correspond to one representative subject and it is possible to observe the delimitation of the functional phases previously referred. The results revealed a large inter-individual variation in the neuromuscular responses.

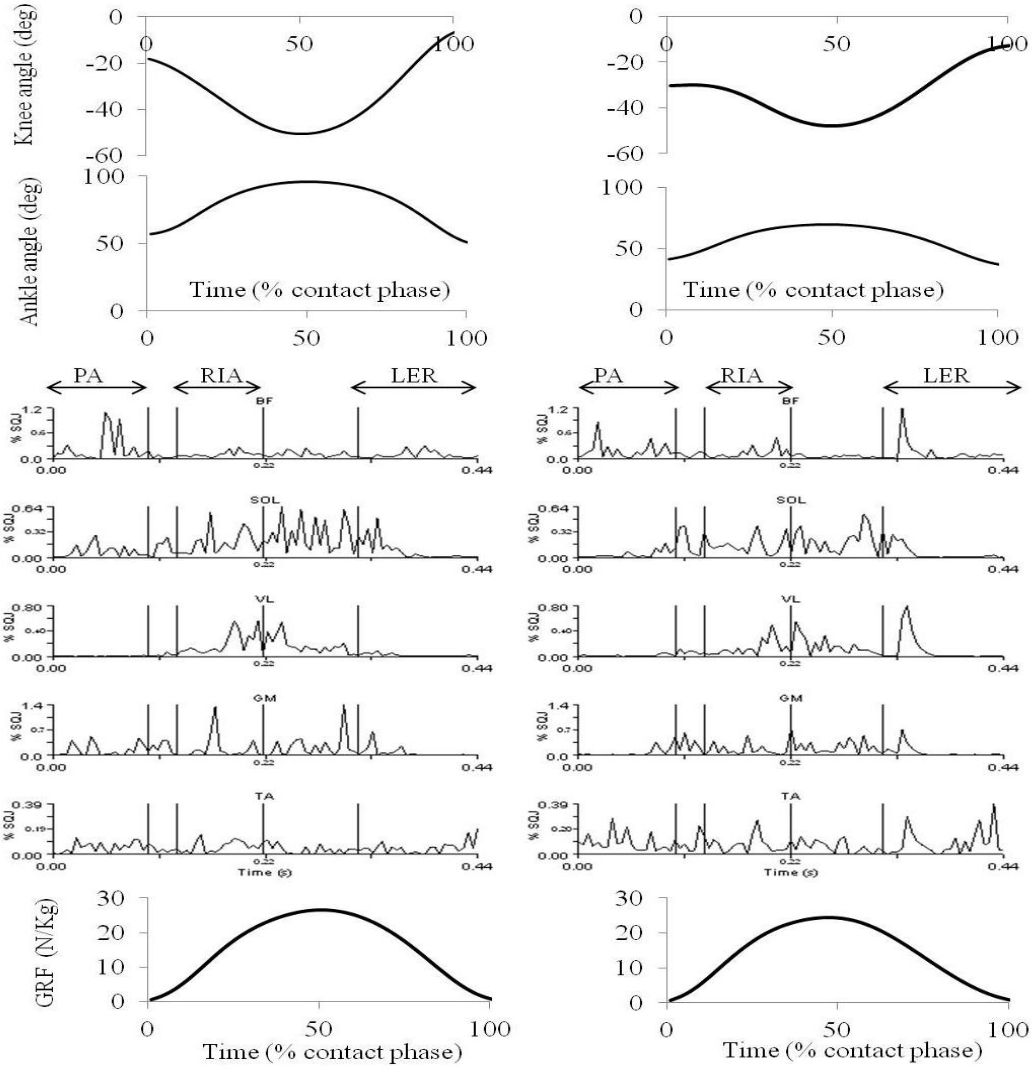

Figure 3-1: Ankle and knee angles, normalized EMG and vertical GRF for two representative hops of one subject. The PA interval corresponds to the pre-activation phase; the RIA interval corresponds to the reflex induced activation phase and the LER interval corresponds to the late EMG response phase. Left column: [0%-10%] hop, right column: [90%-100%] hop.

Regarding pre-activation phase for this subject, there was a small decrease for the SOL and GM with fatigue while the TA slightly increased its activation in that period. In the RIA phase, GM and SOL also decreased while TA and BF showed a small increase. Finally, in LER phase, the TA, BF and VL muscle increased its activation, while both plantar flexors decreased it. The grand mean integrated EMG (iEMG) for each phase and for the ten selected hops for all the subjects are presented in table 3-2.

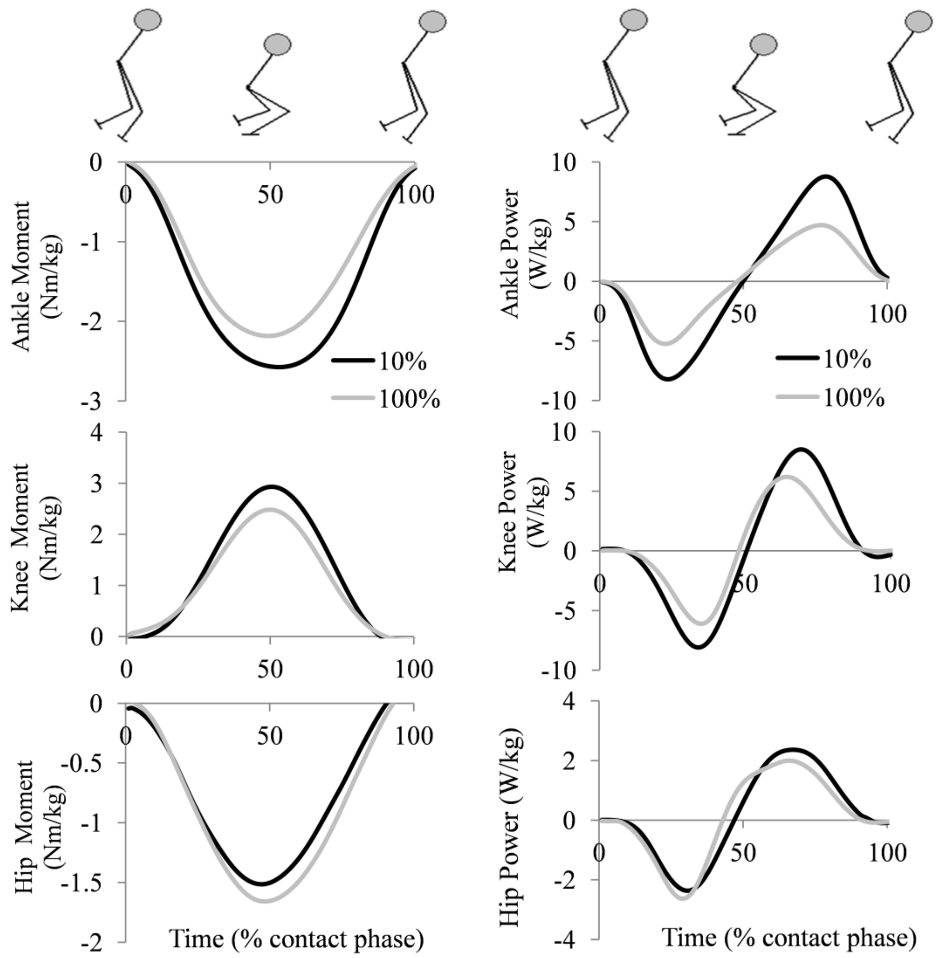

Figure 3-2: Left: Averaged joint moments of force. Right: averaged joint power profiles. Both variables are from the first [0%-10%] (black line) and last [90%-100%] (grey line) selected hops.

The vertical peak of the GRF significantly decreased with fatigue (Figure 3-1), as well as the ankle and knee peak joint moments of force (Figure 3-2 - left). Being a unilateral exercise, it was verified that the abductor/adductor joint moments and joint powers of the hip achieved the highest values in the frontal plane of motion, when compared with the same variables in the sagittal plane. Observing the hip abductor moment we verified an increase with the fatigue condition.

In addition, looking at the joint power profiles, both absorption and propulsion peaks of the ankle and knee joints decreased significantly with fatigue (Figure 3-2 - right) and the hip power peaks didn't reveal significant differences. As a result of fatigue, there was a significant decrease in the ankle and knee joints work, which was accompanied by a significant increase in the work produced at the hip joint, in both braking and propulsion phases of the ground contact (Table 3-1).

The averaged integrated EMG (iEMG) was analyzed during the PA, RIA and LER functional phases and during the braking and propulsion parts of the contact phase, over the ten selected hops (Table 3-2). The results didn't reveal significant differences with the exception of the RIA phase from the BF muscle, which presented an increased activation between the first and the last interval.

Joint stiffness of the hip was the highest, followed by the knee and the ankle. No significant changes were observed in the joint stiffness for the three lower limb joints during the hopping sequence (Table 3-1). Instead, the vertical stiffness revealed a significant increase in the last hop, when compared with the first one, even with a fluctuating pattern during the entire sequence.

Table 3-2: Grand mean of the iEMG (adimensional units) during the PA, RIA and LER functional phases and during the braking and propulsion parts of the contact phase, over the ten selected hops. *significant different from the 10% to the 100% hop. p-value: 0,040.

	SOL					GM					TA				
	PA	RIA	LER	BRK	PRO	PA	RIA	LER	BRK	PRO	PA	RIA	LER	BRK	PRO
10%	0.004	0.012	0.041	0.029	0.026	0.013	0.015	0.045	0.035	0.031	0.006	0.016	0.023	0.030	0.014
20%	0.004	0.013	0.042	0.027	0.030	0.012	0.016	0.054	0.033	0.042	0.006	0.011	0.019	0.018	0.014
30%	0.003	0.019	0.043	0.031	0.032	0.011	0.020	0.042	0.034	0.032	0.007	0.020	0.024	0.032	0.016
40%	0.002	0.016	0.050	0.039	0.028	0.008	0.014	0.045	0.030	0.033	0.004	0.008	0.026	0.020	0.018
50%	0.002	0.015	0.028	0.026	0.019	0.012	0.019	0.029	0.031	0.022	0.004	0.009	0.023	0.020	0.016
60%	0.004	0.019	0.038	0.031	0.028	0.014	0.016	0.043	0.028	0.034	0.005	0.010	0.024	0.018	0.017
70%	0.004	0.020	0.047	0.041	0.030	0.010	0.017	0.046	0.038	0.031	0.005	0.010	0.023	0.018	0.016
80%	0.003	0.018	0.043	0.037	0.026	0.012	0.014	0.035	0.030	0.022	0.004	0.009	0.023	0.020	0.014
90%	0.004	0.022	0.041	0.043	0.026	0.010	0.019	0.038	0.038	0.025	0.005	0.012	0.028	0.024	0.018
100%	0.003	0.015	0.040	0.038	0.022	0.008	0.016	0.042	0.038	0.026	0.005	0.010	0.026	0.023	0.017

	VL					BF				
	PA	RIA	LER	BRK	PRO	PA	RIA	LER	BRK	PRO
10%	0.001	0.014	0.034	0.030	0.020	0.015	0.009	0.028	0.019	0.020
20%	0.001	0.015	0.032	0.032	0.017	0.009	0.009	0.025	0.019	0.019
30%	0.001	0.015	0.030	0.030	0.017	0.017	0.011	0.022	0.024	0.015
40%	0.001	0.016	0.028	0.030	0.017	0.009	0.011	0.023	0.023	0.013
50%	0.001	0.015	0.032	0.030	0.020	0.011	0.010	0.019	0.017	0.015
60%	0.002	0.018	0.032	0.037	0.015	0.008	0.009	0.020	0.020	0.014
70%	0.002	0.020	0.031	0.037	0.017	0.017	0.011	0.035	0.018	0.030
80%	0.002	0.018	0.029	0.034	0.015	0.014	0.009	0.017	0.018	0.009
90%	0.003	0.017	0.037	0.036	0.020	0.014	0.013	0.019	0.024	0.013
100%	0.003	0.016	0.033	0.032	0.019	0.014	**0.012***	0.027	0.022	0.020

3.5 Discussion

The purpose of this study was to investigate the effects of repetitive SSC actions, in the lower limb kinematics, kinetics and neuromuscular behavior. To accomplish that, the subjects performed a unilateral hopping exercise until exhaustion, while the kinematics and GRF synchronized with the EMG of five leg muscles were collected. The fatigue response is very individual and thus, following other studies, the fatigue protocol used was completed to exhaustion to reduce the influence of the inter-subject variability (Nicol, Avela, & Komi, 2006).

The observed results regarding the joint kinematics at the instant of the ground contact are in accordance with Self & Paine (Self & Paine, 2001) since the subjects, with fatigue, landed with the knee more flexed and the ankle more dorsi flexed. This may be a mechanism of compensation of the knee to spare the fatigued ankle. Therefore, the limited plantar flexed position at ground contact when the subject is fatigued is compensated by an increase in the knee flexion to help absorb the forces during impact. On the other hand, the decrease of the vertical GRF peak with the accumulation of fatigue could be an indicator of a safer landing strategy (Dufek & Bates, 1990). The drop in GRF is probably an indicator of reduction in tolerance to repeated stretch-loads with the increased fatigue (Komi, 2000).

Joint powers can help us to describe the mechanisms by which kinetic energy is dissipated during ground contact (Weinhandl, Smith, & Dugan, 2011). Both ankle and knee power peaks decreased significantly with fatigue. Although the hip power remains unchangeable, previous research showed that, with fatigue, the hip moment contribution to accelerate the center of mass of the body during hopping increases, while the ankle and knee joint moments of force decrease their contribution (João & Veloso, 2013). Moreover, our results reveal a significant decrease in the ankle and knee moments of force, while the hip moment remains the same. Other investigations refer that an active trunk flexion during landing can alter lower extremity kinematics in a way that may reduce knee injury risk. In fact, a study conducted by Kulas and colleagues (Kulas, Hortobagyi, & Devita, 2010) reported that subjects who actively flexed the trunk during landing were able to use more the hamstring muscles, therefore decreasing the load in the knee extensors and reducing the knee anterior shear forces. Actually, in the current study, the BF muscle significantly increased its activation in the RIA phase in the fatigue condition, which can be a mechanism used to control and reduce the load in the knee extensors.

Regarding the neuromuscular functioning, a non-fatiguing SSC is characterized by a low EMG activity in the concentric phase of the cycle and a greater contribution of the stretch-reflex but this behavior is dependent in the type of hopping, level of the applied load and level of fatigue that is achieved. With exhaustive SSC fatigue, the reflex contribution decreases significantly (Komi, 2000). Our results didn´t show significant differences in the EMG integrated over the functional phases, namely PA, RIA and LER. Nevertheless, the rectified EMG is analyzed at an individual level (Figure 3-1) and substantial changes could be observed in the RIA phase: the EMG bursts are consistent with the suppression of reflex thus supporting the lack of power observed at the ankle and knee joints. In SSC contractions, the reflexes are important by making the force output more powerful, due to the efficient transition from the pre-activated and stretched muscle-tendon complex to the concentric phase of the hopping activity (Komi, 2000). With fatigue, this transition may become less efficient, leading to an immediate reduction in force and reflex sensitivity (Kuitunen et al., 2002) but again, it is a subject-specific neuromuscular response. Other studies with SSC showed that no changes occurred in EMG activity patterns during drop jumps (Kyrolainen et al., 2005) or in plantar flexor muscles EMG during pre-landing and eccentric phase of vertical jumps after SSC exercise (Kubo et al., 2007). Neuromuscular fatigue might be a contributing factor in musculoskeletal injuries during prolonged exercise and there is evidence that it changes landing postures and landing compensatory strategies (Weinhandl et al., 2011).

Repeated SSC muscle actions can modify the mechanical properties of the muscle tendon unit, which are time- and history-dependent due to the viscoelastic nature of the tissue (Nicol et al., 2006). In a repeated SSC action such as hopping, we found no differences in joint stiffness but a significant increase in vertical stiffness was verified from the first to the last selected hop, although some fluctuations occurred during the entire sequence. Vertical stiffness was calculated through the quotient of the vertical GRF peak by the displacement of the body's center of mass from the instant of the contact until the minimum height. In the last hop both variables decreased but in a proportion where the CM displacement decreased more than the GRF peak, denoting an increase in vertical stiffness. Nevertheless the lack of changes in joint stiffness, a previous study from our group showed that the modulation of the ankle stiffness was the primary mechanism for adjusting leg stiffness in hopping using a virtual passive controller as the leg spring (Kim et al., 2013). Moreover, leg stiffness increased in some subjects and decreased in others, according to the activation pattern observed in the main muscles acting about the joints of the leg. Thus the mechanical action with the fatigue state is highly subject-dependent, particularly in the case of non elite athletes due to their low tolerance to high demanding loads. We should also consider the fact that vertical or leg stiffness during a jump can be altered by verbal instructions (Arampatzis et al., 2001).

In conclusion, using the unilateral hopping model it is possible to study the fatigue phenomena and observe the alterations in the kinematics, kinetics and neuromuscular pattern of the lower extremity. The joint power data seems to be clear regarding the loss of power in the ankle and knee joints. These results are in agreement with the reduction in the EMG bursts of the plantar flexors (Figure 3-1) and expected for an exhaustive SSC. At the same time, the BF muscle increases its activation (Table 3-2), probably to reduce the load in the knee extensors.

3.6 Conflict of interest statement

The authors have no personal or financial conflicts of interest related to publication of the present work.

3.7 Acknowledgments

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Chapter 4

4 Quantifying the effect of plyometric hopping exercises on the musculoskeletal system: contributions of the lower limb joint moments of force to ground reaction forces in hopping exercise[2]

[2] Published as:

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4.1 Synopsis

The purpose of this study was to estimate the ability of joint moments of force to transfer mechanical energy through all the leg segments during a cyclic hopping sequence, performed until exhaustion. The technique was applied to data from 4 healthy active students to characterize the relative contribution of the lower limb net joint moments of force to accelerate the ankle, knee and hip joints. Our findings show that the strategies used to maintain the same jumping height rely on the balance between the net joint moments to guarantee the acceleration of the joints. It seems that while the ankle and knee moments reduce their contribution to accelerate the ankle and the knee joints, the hip moments increase their participation and have an important influence in the re-arrangement of the musculoskeletal system to maintain the same mechanical output.

4.2 Introduction

The knowledge of the exact role of a muscle in a particular movement is an important goal of the majority of researchers in the contexts of biomechanics, rehabilitation, performing arts, sports, etc. Muscles have critical participations in the performance of a motor task. As Zajac (2002) mentioned, «muscles redistribute the net mechanical energy of the body segments because each muscle force causes reaction forces throughout the body with the net effect being to accelerate some segments and decelerate others». This means that a muscle acts in all joints and segments, even joints it does not span and segments to which it is not attach to (Zajac, 1993). This phenomenon, also called induced acceleration could be a precious tool to analyze in greater detail and directly quantify a muscle/muscle group contribution to a specific motor task. The magnitude of these accelerations is a function of both the intensity of the muscle forces as well as the displacement of the joints (position and orientation of the body segments).

Analyses of induced accelerations (IAA) have been mainly used with normal and impaired gait (Goldberg & Kepple, 2009; Riley & Kerrigan, 1999; Siegel, Kepple, & Stanhope, 2006, 2007), in order to study the changes in muscle contributions to motion. However, most dynamic human movements in sports activities involve cyclic stretch-shortening actions with maximal mechanical power production, so this technique should be of benefit in analyzing the different contributions of lower limb muscle groups over a time period during cyclic exercise.

Moreover, this tool can be useful to assess muscle groups adaptive compensations when muscle fatigue occurs. It is especially useful to understand changes in coordination that may occur even if the athlete is able to maintain the same mechanical output, for instance the same jumping height.

Jumping has been studied at diverse levels such as in vivo mechanical properties (Fukashiro, Hay, & Nagano, 2006; Kawakami, Muraoka, Ito, Kanehisa, & Fukunaga, 2002; Kurokawa, Fukunaga, & Fukashiro, 2001; Lichtwark & Wilson, 2005) or modeling and simulation (Anderson & Pandy, 1993, 1999; Delp et al., 2007; Delp et al., 1990; Nagano, Komura, Yoshioka, & Fukashiro, 2005; Neptune, Kautz, & Zajac, 2001; Neptune, Zajac, & Kautz, 2004; Pandy, 2001; Zajac, 1993). The ability to perform a vertical jump is a crucial skill, especially for sports such as volleyball, basketball or football. As Rodacki, Fowler, & Bennet (2002) refer, the execution of this motor task depends on the coordination of two important aspects: 1) the segmental actions of the body and 2) the net joint moments that have to be generated to achieve the required mechanical task. So that is why jumping biomechanics has its relevance since plyometrics are common exercises used to improve speed and muscular power in athletes. Various vertical jumps can be used as a possible approach to study the force-length relationship in the human muscle-tendon complex (Bobbert, Gerritsen, Litjens, & van Soest, 1996; Gollhofer, Strojnik, Rapp, & Schweizer, 1992). The drop jump is one of the most common plyometric exercises, where the stretch-shortening muscle actions are intensified (Weinhandl, Smith, & Dugan, 2011).

In a stretch-shortening cycle, the viscoelastic characteristics of the musculo-tendon complex are extremely important to guaranty the effectiveness and efficiency of human performance (Fukashiro et al., 2006). A predictor model for this type of behavior is quite relevant since stress and strain injuries are commonly linked with these muscle actions. The limits of performance during prolonged exercise have been subjected to several studies. Fatigue is associated with the inability to maintain the same level of mechanical output, and several causes are associated. It is being clearly showed by several authors (Goldberg & Kepple, 2009; Kepple, Siegel, & Stanhope, 1997; Siegel et al., 2007) that the effect of a muscle group action or a joint moment of force can accelerate body segments that are located far on the segmental chain through a coupling effect. The relative contribution of the joint moments of force on the GRF or on joint angular acceleration can be estimated using induced acceleration analysis.

These principles of induced acceleration lie on the fact that the moments produced by muscle forces around a joint will accelerate all joints of the body and not just the joints they cross. Kepple (1997) found that the forward progression in gait was produced primarily by the ankle plantar flexors and in some circumstances with the help of the knee extensors. During single limb support phase, the upper body was supported mainly by the plantar flexors action, while during double limb support it was the sum of the plantarflexors, knee and hip extensors actions. Zajac & Gordon (Zajac & Gordon, 1989) demonstrated that the gastrocnemius may sometimes be a knee extensor, although anatomically is considered a knee and ankle plantarflexor.

4.1.1 Governing Equations of Motion

The dynamic equations of motion for a certain task can be derived using different methods. The most common is the Newton-Euler method (Orin, Mcghee, Vukobratovic, & Hartoch, 1979) where the body segments are built through free-body diagrams, where the external forces and torques that are acting on each segment are shown. This form of interpretive analysis of the movement is based on the equations of motion for a two segment planar pendulum (Figure 4-1).

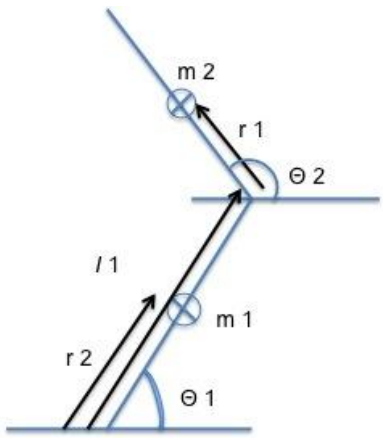

Figure 4-1: Two segments planar pendulum model, where “m” is the mass of the segment, r is the distance between the center of mass of the segment and the rotational point (joint), I is the inertia of the segment and θ is the joint angle.

From here, net joint moments τ_1 and τ_2 are expressed as:

(Eq.4-1)

(Eq.4-2)

These equations can be expressed in a matrix form:

(Eq.4-3)

or also in the following form:

(Eq.4-4)

Where is the joint accelerations matrix, is the inverse inertia matrix (where the segments inertial parameters and center of mass positions are taken into account), is the joint moments matrix, is the Coriolis terms matrix, is the Gravitational terms matrix and is the external forces matrix. Setting , and terms to zero allows us to obtain the accelerations produced only by the joint moments (Kepple et al., 1997).

With the ground reaction force data experimentally collected and the motion capture data, as well as the inertia parameters, the net joint moments of force are calculated trough inverse dynamics (see section 4.2.3). To calculate the induced accelerations, these experimental net joint moments and the segments kinematic data collected from the motion capture serve as input to equation 4-5. These instantaneous accelerations are calculated given one of the input net moments ( ) and the position and orientation of the segments. The output is the acceleration of a joint or segment's center of mass.

(Eq.4-5)

To isolate the contribution of one particular joint moment to the acceleration of all the joints of the model implies the assumption that those other joints have moments and stiffness equal to zero. Accelerations' magnitude is dependent not just on the moment magnitude but also on the configuration of the body segments.

Therefore, the purpose of our study was to use a similar IAA to estimate the relative contribution of lower limbs' joint moments to transfer mechanical energy through all the leg segments during a cyclic hopping exercise. We also wanted to understand how this energy transfers changes due to specific fatigue, as a consequence of changes in muscle coordination. The technique was applied to data from 4 subjects to characterize the relative ability of the lower limb joint moments to accelerate the ankle, knee and hip joints (Kepple et al., 1997; Siegel et al., 2006, 2007).

4.2 Measurements and Analysis

4.2.1 Experimental procedure

Four healthy recreationally active students (weight: 62.4±5.3 kg and height: 1.7±0.1 m) performed a sequence of unilateral jumps with their dominant lower limb until exhaustion, being exhaustion defined as the inability to maintain a prescribed mechanical output. To establish a control parameter for the jump height, a squat jump (SQJ) was performed prior to the hopping task. The minimum height of all jumps from the hopping sequence was 60% of the maximum height achieved in the SQJ. The subjects performed the hops on a contact mat placed on the top of a force plate, facing a computer monitor where they could see the feedback of their jumping height. They were instructed to keep their hands on their waists to minimize arm motion because the head, arms and trunk were modeled as a single segment (Kepple, 1997).

Task failure was considered when the subjects couldn't achieve the minimum required height. It was considered from each subject an indicative jump from each 10% interval. The first and last selected jumps were chosen for the analysis, to determine the effects of a repetitive hopping sequence on the induced acceleration results.

4.2.2 Motion capture

Motion capture was collected with 10 cameras Qualisys (model: Oqus-300) operating at 200Hz. 24 reflective markers and 4 marker clusters were placed in predefined anatomical protuberances (acromial, stern, anterior-superior and posterior-superior iliac spines, lateral and medial femoral condyles, lateral malleoli of the fibula, medial malleoli of the tibia, 1st and

5th metatarsal heads and one marker cluster for each thigh, shank and foot segments), as can be seen in Fig. 4-2A. These markers were used for the reconstruction of eight body segments (trunk, pelvis, right and left thighs, right and left shanks and right and left feet) in Visual 3D software (Visual 3D Basic RT, C-Motion, Inc., Germantown, MD) (Fig. 4-2B).

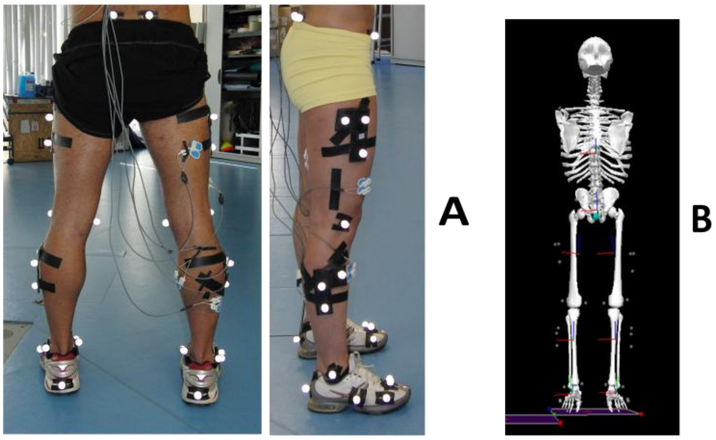

Figure 4-2: A) Segments' marker setup (two pictures on the left) and B) segment's model reconstruction in Visual 3D (picture on the right)

Each body segment is reconstructed as a rigid body, with a length and an embedded coordinate system. Visual 3D assumes that each segment moves accordingly with the motion of the tracking markers that are attached to it. For more detail with this methodology, please see Appendix A from (João, Amado, Veloso, Armada-Da-Silva, & Mauricio, 2010). Regarding this, the system tracks the position and orientation of each segment, allowing for the calculation of all the kinematic variables. For the kinetic data calculation, segment's mass is based on Dempster's anthropometric data (1955) and moments of inertia properties are based on Hanavan's study (1964).

Ground reaction force (GRF) was collected with a Kistler force plate (type: 9865B). For this analysis, computed variables included spatial-temporal variables, joint angular displacement, joint velocities and net internal joint moments' contribution to ground reaction force. They were all calculated in Visual 3D software. The results were processed with a lowpass 4th order Butterworth filter with a frequency cut of 6 Hz.

4.2.3 Induced acceleration analysis

The net joint moments that serve as input to the IAA computation were firstly calculated by inverse dynamics. This means that, the ground reaction force obtained experimentally as well as the motion capture data and inertial parameters of the segments were used to calculate the joint moments. Inverse dynamics is a process that computes the net effect of all the anatomical structures across a joint (especially muscles and ligaments) to produce a certain motion of the joint. To perform these computations, the Newton-Euler equations are used:

(Eq.4-6)

(Eq.4-7)

where the force ( ) equals mass ( ) times linear acceleration ( ) and moment of force ( ) equals moment of inertia ( ) times angular acceleration ( ). From equations 4-6 and 4-7, inverse dynamics derives the moment of force based on the segment's or joint's motion.

Now each of the joint moment previously computed was used to calculate the induced acceleration at that particular joint. The contribution of each joint moment to the angular acceleration was calculated using Visual 3D's IAA Module. All the procedures regarding the technical aspects of this software are refereed in the C-Motion Wiki Documentation page that can be found in the following URL: http://www.c-motion.com/v3dwiki/index.php?title=Main_Page.

Each subject's joint and segment positions were obtained from the motion capture and served as input to the model. Gravity and all joint moments were set to zero. One joint moment then was entered into the model and the resultant joint angular acceleration was computed (Kepple et al., 1997). The input joint moment was then set back to zero, and another joint moment or gravity was sequentially entered into the model.

The model output provided the portion of the hip, knee and ankle joint angular accelerations that were generated individually by each input joint moment. Ankle, knee and hip joint induced angular accelerations were computed (Kepple et al., 1997; Siegel et al., 2006, 2007). The IAA was performed at all frames of the ground contact phase of each jump. The ankle was designed as a universal joint (plantar/dorsi flexion and inversion/eversion), the knee as a revolute joint (flexion/extension (flex/ext)] and the hip as a spherical joint [flexion/extension, abduction/adduction (abd/add) and internal/external (int/ext) rotation].

4.3 Results

Concerning temporal variables of the 4 subjects' jumps (Table 4-1), it could be seen in the final hops that, as fatigue increased, the subjects wanted to take their feet off the ground the fastest as they could, therefore decreasing the stance phase and the braking phase.

Table 4-1. Temporal variables of the four subjects' jumps.

Temporal variables (mean ± SD)	Subj 1	Subj 2	Subj 3	Subj 4
Contact phase duration Start hop (s)	0.391±0.016	0.316±0.014	0.339±0.015	0.413±0.027
Contact phase duration Final hop (s)	0.362±0.017	0.308±0.011	0.198±0.018	0.470±0.029
Suspension phase duration Start hop (s)	0.225±0.012	0.242±0.024	0.246±0.022	0.242±0.025
Suspension phase duration Final hop (s)	0.253±0.211	0.162±0.016	0.256±0.022	0.165±0.08
Braking phase duration Start hop (s)	0.181±0.011	0.155±0.010	0.170±0.011	0.197±0.011
Braking phase duration Final hop (s)	0.161±0.011	0.134±0.009	0.092±0.011	0.210±0.009
Propulsion phase duration Start hop (s)	0.209±0.009	0.161±0.005	0.175±0.007	0.216±0.025
Propulsion phase duration Final hop (s)	0.201±0.010	0.174±0.006	0.106±0.009	0.260±0.018

Both suspension phase and propulsion phase duration do not change in the same way in the 4 subjects, which may be explained by the changes observed in the hopping pattern with an increase in the jumping velocity and a less controlled movement demonstrated by some of the subjects.

Joint angles, velocities and internal net moments (calculated by inverse dynamics) from the first and the last hops are presented in Fig.4-3. In a general way, there is a decrease in the angular amplitude and angular velocity flexion and extension peaks of the joints (Fig.4-3A). Joint moments also decreased, especially ankle moment, with the exception of the hip flex/ext moment that increased. These results could be explained by an adaptation of the system to maintain the same mechanical output. Interestingly, both abd/add and int/ext rotational moments have a smoother pattern when compared with the flex/ext moment (Fig.4-3B).

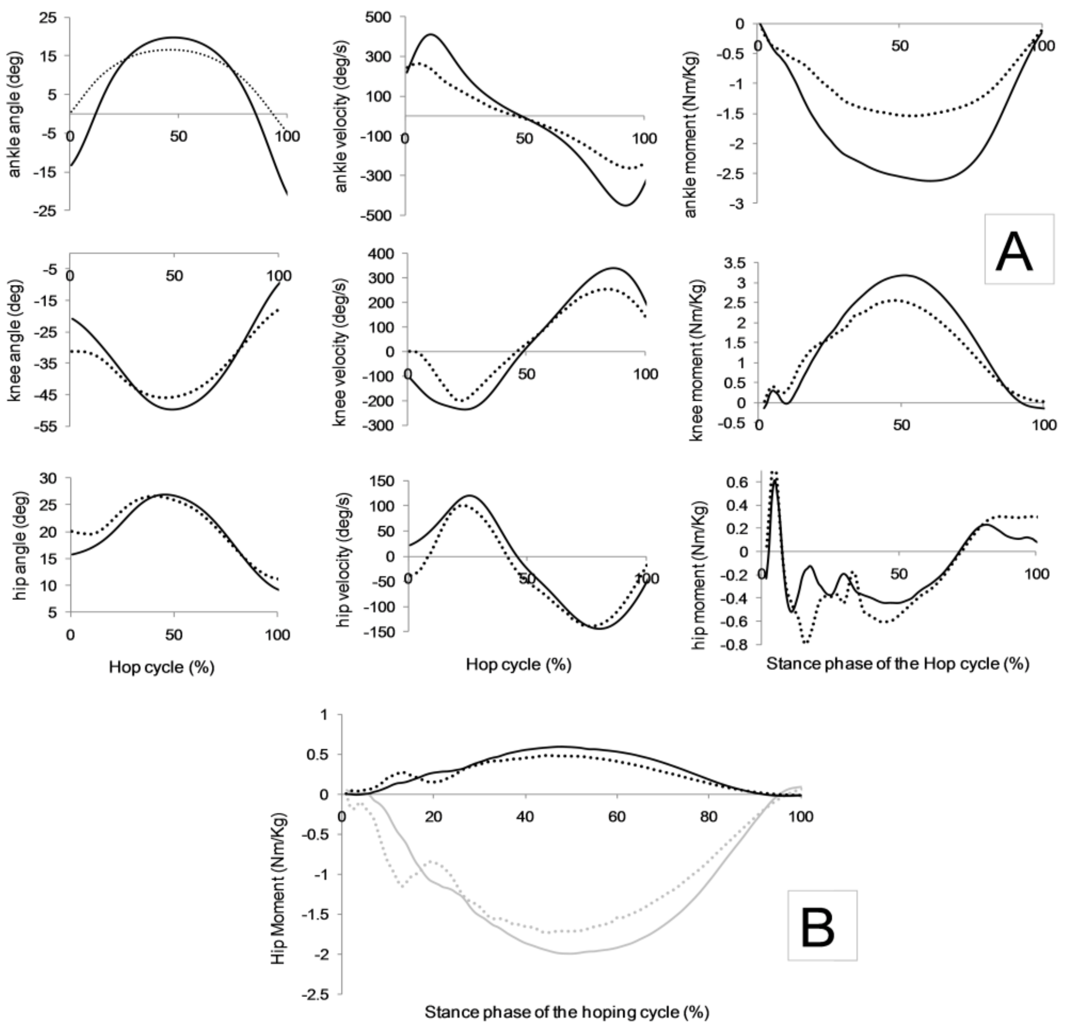

Figure 4-3: A) Plots of angular displacement, angular velocity and net joint moments (calculated by inverse dynamics) from the right ankle (line 1), right knee (line 2) and right hip (line 3) during the first (solid line) and last (dotted line) hopping jump. All the variables concern to the sagital plane of motion. B) Abduction/adduction joint moment (black line) and internal/external rotation moment (grey line) of the right hip joint. First hop sequence (solid line), last hop sequence (dashed line).

The joint angular accelerations and GRFs induced by the ankle, knee and hip flex/ext moments are reported in Figs.4-4, 4-5 and 4-6. Analyzing joint by joint, the ankle was highly accelerated by the knee and ankle moments during braking phase reaching a peak of 600 and -800 deg/s^2, respectively. The hip contribution remains the same although it increases in the final hops with a peak value of 225 deg/s^2 right after the foot contact. The abd/add and rotational moments of the hip did not contributed significantly to the ankle flex/ext acceleration (results not shown) but they accelerated the ankle into inversion/eversion and rotation more substantially in the first hops. The three joint moments contributed to the knee acceleration, in particularly the ankle and knee moments (peak value of -500 deg/s^2 and 650 deg/s^2, respectively). Concerning hip acceleration, again ankle and knee moments were the main contributors reaching peak values of 400 and -400 deg/s^2, respectively.

Each moment contributed to accelerate the joints but, with repetition, that contribution decreased for the ankle and knee moment and remained more or less constant for the hip flex/ext moment. However, in the final jumps hip moment increased its contribution to accelerate the ankle, therefore compensating the decrease of the ankle and knee joint moments contribution to maintain the jumping height (performance criterion).

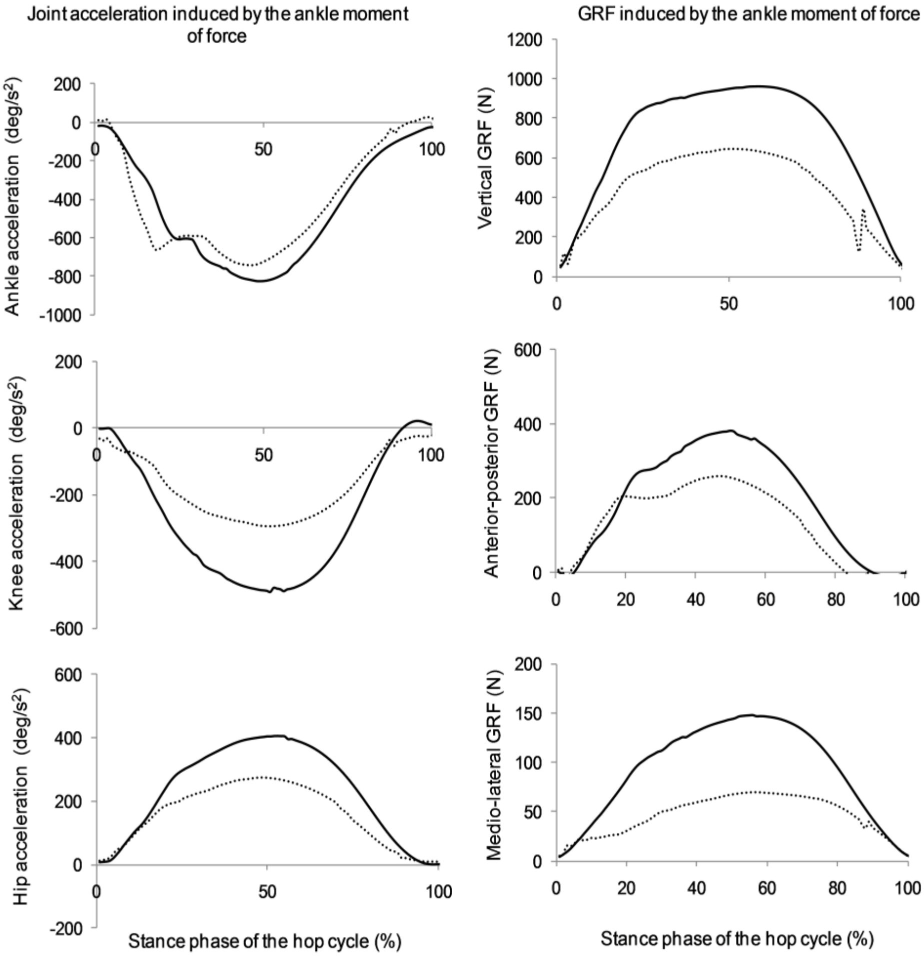

Figure 4-4: Joint angular accelerations and GRF induced by the ankle internal net moment. First hop sequence (solid line) and last hop sequence (dotted line). The first column corresponds to the ankle, knee and hip joint's acceleration and the plots of the second column refers to the three components of the GRF that are induced by the same ankle moment. All the joint acceleration variables concern to the sagital plane of motion.

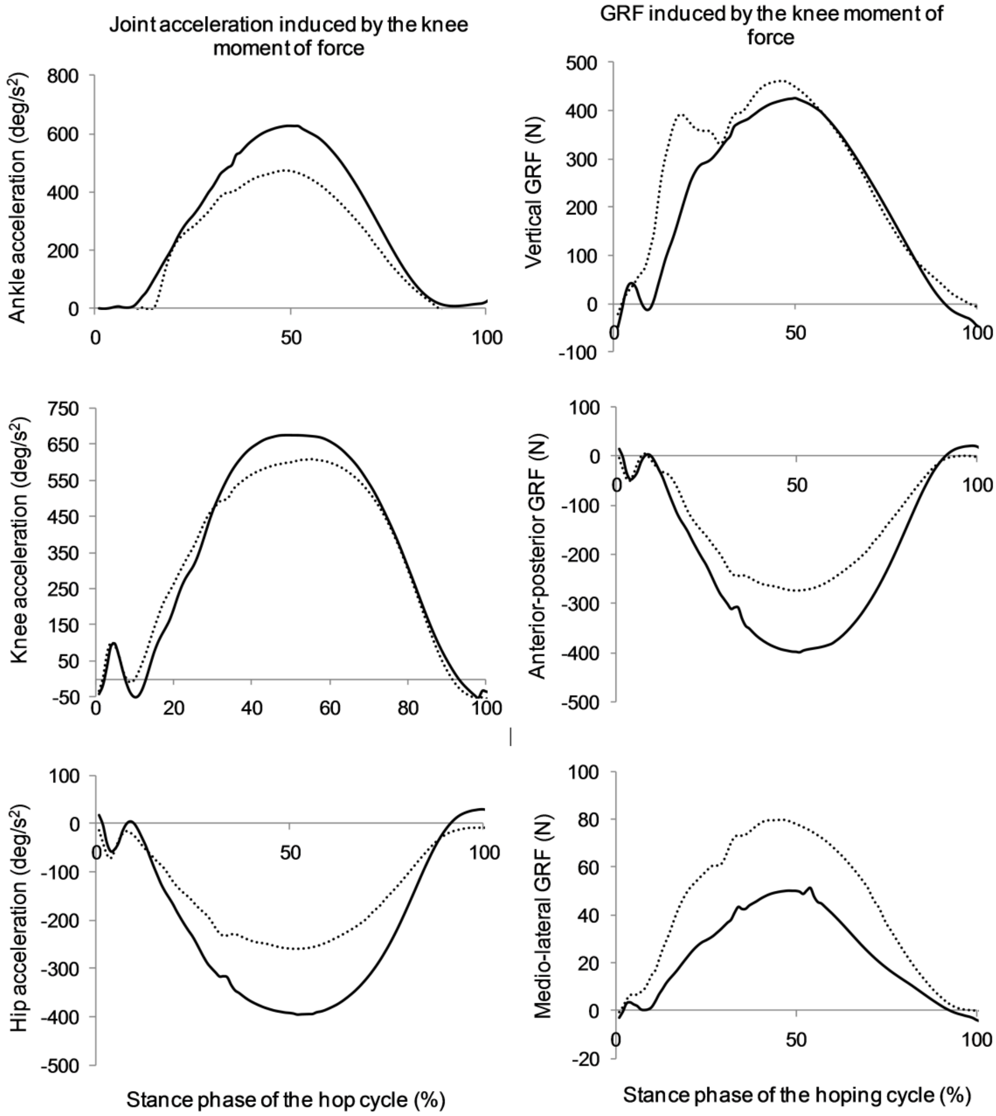

Figure 4-5: Joint angular accelerations and GRF induced by the knee joint moment. First hop sequence (solid line) and last hop sequence (dotted line). The first column corresponds to the ankle, knee and hip joint's acceleration and the plots of the second column refers to the three components of the GRF that are induced by the same knee moment. All the joint acceleration variables concern to the sagital plane of motion.

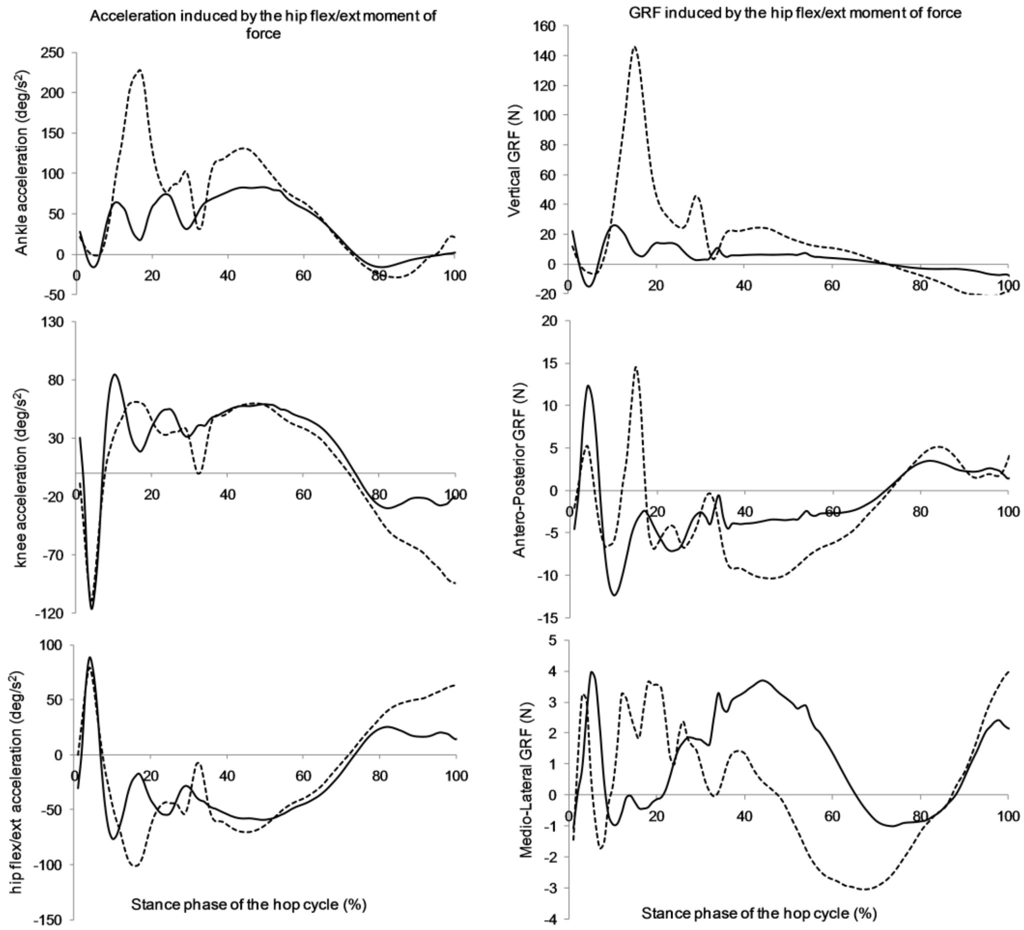

Figure 4-6: Joint angular accelerations and GRF induced by the hip flexor/extensor moment. First hop sequence (solid line) and last hop sequence (dotted line). The first column corresponds to the ankle, knee and hip joint's accelerations induced by the hip moment and the plots of the second column refers to the three components of the GRF that are induced by the same hip moment. All the joint acceleration variables concern to the sagital plane of motion.

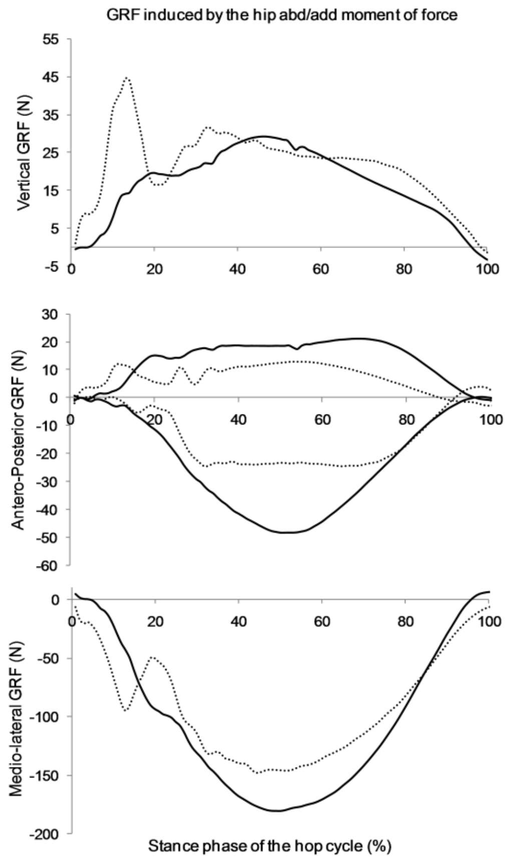

Figure 4-7: GRF induced by the hip abductor/adductor rotational moment. First hop sequence (solid line) and last hop sequence (dotted line).

Looking at the vertical component of the GRF induced by each joint moment, the ankle induced nearly 1000 N, the knee moment contributed 400 N and hip moments had a lesser contribution. A clear contribution of the ankle and knee joint moments was observed in GRF, which decreased along the jumping task and it were observed two patterns for the contribution of the hip abd/add moment: some subjects presented a positive curve pattern whereas others presented a negative curve, both decreasing in time (Fig.4-7). This effect may be due to the jumping technique adopted by each subject, pushing the floor in opposite directions to maintain their bodies always in the same position. The left limb (swing limb) joint moments contribution to the joint accelerations was also calculated, but the results are residual when compared with the right lower limb, so they were not reported here.

4.4 Discussion

The magnitude of the accelerations depend both on the magnitude of the muscle forces as well as the displacement of the joints. Patients with the loss of function in one or more muscle groups will use adaptive strategies to use other muscle groups to accelerate the joints they do not cross. This is one way to produce compensatory mechanisms that enable the subject to perform a certain motor task. Besides the clinical relevance of this type of analysis, this approach should also be addressed to fatiguing mechanical load tasks, such as jumping exercise until exhaustion. In the same way, the musculoskeletal system will use a compensatory strategy to avoid task failure.

Our findings showed that the strategies used to maintain the same jumping height rely on the balance between the joint net moments to guarantee the acceleration of the joints. While the ankle and knee moments reduced their contribution to accelerate the ankle and the knee joints, the hip moments increased their participation. This distal to proximal redistribution of the joint moment's contribution to accelerate the joints can be a mechanism originated by the fatigue condition. The hip rotational moment contribution to the vertical GRF is negative. A possible reason for this is the fact that even with the plantarflexors very active (and consequently producing a positive GRF), the subjects simultaneously flex their hip, and the hip flexors would reduce the GRF produced by the plantarflexors muscle group (producing a negative induced GRF). This effect was even marked in the final hops, where it could be seen a change in mechanical performance and a slight drop of force after impact, probably related with the loss of tolerance to the repeated stretch load as fatigue progresses (Komi, 2000).

Fatigue leads to failure after repeated application of stresses, and changes in tendon elastic properties may have a positive effect on the dynamics of the muscle-tendon complex during stretch-shortening cycle exercise. Muscle force generating ability is compromised by the tendinous structure compliance, since it influences the length and shortening velocity of the muscle, according to the force-length-velocity relationship. (Lichtwark & Wilson, 2005) studies with Achilles tendon show that the majority of the strain occurs in the tendon compared with the contractile component. If the tendon experiences an elevated elongation, the muscle component may be less affected and could actuate at a lower shortening velocity. Benefiting from the force-velocity relationship, gastrocnemius would then be able to produce more force.

In vertical jumping the hip is commonly assumed to be a pin joint which has only one degree of freedom because sagital plane is assumed to dominate the performance of the motor task (Zajac & Gordon, 1989). However, in this case of unilateral vertical jumping (a task that also implies the maintenance of balance), the hip joint was considered a spherical joint and has an important influence in the re-arrangement of the musculoskeletal system to maintain the same mechanical output (as it was found in the medio-lateral GRF induced by the abd/add hip moment).

Fatigue caused by stretch-shortening cyclic exercise normally causes reversible damages in the muscles and alters muscle and joint's mechanics and stiffness (Komi, 2000). The kinematic changes of a fatigued athlete are the result of muscle performance impairments that contribute to the athlete's inability to maintain the same mechanical output for a long period of time. Those changes may increase injury risk potential, especially in athletes with less experience or muscle imbalances (Gerlach et al., 2005). Most coaches agree that plyometric training is a good choice if we want to improve the athlete's jumping height although some discrepancy exists in the literature (Markovic, Jukic, Milanovic, & Metikos, 2007). It is also a common way to assess muscle power capabilities.

Regarding this, it would be of both scientific and practical relevance to quantify the effect of plyometric training or plyometric exercises until exhaustion on vertical jump ability. Foot impact with the ground during athletic activities can inflict powerful loads on the musculoskeletal system. When fatigue increases the musculoskeletal system doesn't respond in the same way and the focus of our training program needs to be re-arranged due to those changes. To prevent musculoskeletal injuries caused by jumping exercises it is required an appropriate training program. Knowing that a fatigue condition is expected to occur and may cause impairments in joint mechanics, training programs should prepare the body to resist to fatigue and maintain the same mechanical output (for instance, a perfect technique). In order to achieve this, it's extremely important to understand how the muscles and tendons change their ability to perform work when the musculoskeletal system is fatigued.

When we perform an analysis of the data that we collect in our laboratories, we may do it more in a descriptive way, in an interpretative way or in a predictive way. Depends on what is our aim. In this study, Induced acceleration analysis can be seen as an interpretive method to study the contributions of each lower limb joint moment to vertically accelerate the body during a hopping exercise. The interpretation of the data using IAA may be included between the forward dynamics and the inverse dynamics methods because it can use as input both experimental as simulated data. The explanation of motion data is a significant step in the regular use of motion analysis. This interpretation process allows the data to be presented to the clinician or to the coach in a more comprehensible way, particularly the use of adaptive strategies that rely on the ability of different muscles or muscle groups to accelerate the joint which they not span. This analysis has recently been in the center of attention of many investigators in the biomechanics community (Hamner, Seth, & Delp, 2010; Liu, Anderson, Pandy, & Delp, 2006; Liu, Anderson, Schwartz, & Delp, 2008).

4.5 Conclusions

Induced acceleration analysis is a technique that enables the quantification of the contribution of individual joint moments, muscle forces or gravity to the joint angular accelerations, GRF or segmental powers in a specific task. However, these contributions are still not well defined and are sensitive to the model formulation and representation (Chen, 2006). Our study uses this method to understand cyclic unilateral hopping exercise until exhaustion, thus providing us useful hints concerning musculoskeletal adaptations in this specific task.

Our results showed a change in intersegmental coordination where the relative contribution of ankle plantarflexor moments and knee extensors moments seem to decrease while hip extensor moment increases. This change could not be observed using a contact mat or even a force plate. These results indicate that plantarflexors reach exhaustion and could be prone to injury. Moreover, the training specificity diminishes considering that, for running, plantarflexors seem to have a strong contribution to both horizontal and vertical acceleration of the center of mass, for instance (Hamner et al., 2010).

One of the limitations of this study is the use of joint moments instead of muscle forces or muscle excitation patterns as the input for the IAA. That analysis should also be addressed in this kind of motor tasks, in order to better understand how musculoskeletal structures adjust to maintain the required mechanical output. The knowledge of this type of behavior could be extremely helpful for injury prevention and proper exercise progression.

4.6 Acknowledgements

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Chapter 5

5 Synergistic interaction between ankle and knee during hopping revealed through induced acceleration analysis [3]

[3] Submitted as:

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5.1 Abstract

The forces produced by the muscles can deliver energy to a target segment they are not attached to, by transferring this energy throughout the other segments in the chain. This is a synergistic way of functioning, which allows muscles to accelerate or decelerate segments in order to reach the target one. The purpose of this study was to characterize the contribution of each lower extremity joint to the vertical acceleration of the body's center of mass during a hopping exercise. To accomplish this, an induced acceleration analysis was performed using a model with eight segments. The results indicate that the strategies used during a hopping exercise rely on the synergy between the knee and ankle joints, with most of the vertical acceleration being produced by the knee extensors, while the ankle plantar flexors act as stabilizers of the foot. This synergy between the ankle and the knee is perhaps a mechanism that allows the transfer of power from the knee muscles to the ground, and we believe that in this particular task the net action of the foot and ankle moments is to produce a stable foot with little overall acceleration.

5.2 Introduction

Synergies occur in nature as an organization of elements which have the role of maintaining the stability properties that is, the low variability of a certain performance variable (Latash, Gorniak, & Zatsiorsky, 2008). Muscles can generate force and develop mechanisms to exchange energy among segments, whether they are performing isometric, concentric or eccentric actions (Zajac, 2002; Zajac, Neptune, & Kautz, 2002). Muscles can work in both functional and synergistic ways, that is, they generate and harmonize the force so individual segments gain and loose energy according with the type and aim of the mechanical task (Zajac, 2002). The forces produced by the muscles can deliver energy to a target segment they are not attached to, by transferring this energy throughout the other segments in the chain. This is a synergistic way of functioning, which allows muscles to accelerate or decelerate segments in order to reach the target one (Zajac, 2002).

Induced acceleration analysis (IAA) is a technique based in the dynamic coupling effect caused by the multiarticulated nature of the body (Zajac, 1993, 2002; Zajac & Gordon, 1989). Dynamic coupling means that when a muscle contracts it produces acceleration, not only in those segments that are spanned by that muscle but on all body segments of the chain, due to the intersegmental forces. Thus, this technique allows the direct quantification of a joint moment contribution to the acceleration of each joint of the body and to the body center of mass.

Simple multijoint motor tasks such as vertical jumps have been used to study this intersegment coordination and synergistic behavior. Bobbert, Huijing, & Schenau (1986) studied the behavior of the human triceps surae muscletendon complex during plantar flexion in unilateral jumping and concluded that the biarticular action of gastrocnemius increases significantly the jumping performance. Gregoire, Veeger, Huijing, & van Ingen Schenau (1984) reported that, at the end of the push-off of vertical jumps, the hip and knee joints showed high extension velocities and the power delivered by the monoarticular extensors of the hip and knee joints was carried to the distal joint (ankle) through the biarticular muscles. Simultaneously, the ankle joint showed high plantar flexion velocity, meaning that the biarticular muscles crossing these joints were able to contract at a relatively low speed. As a consequence, biarticular muscles would be able to produce force at a lower contraction velocity, thus generating higher forces, and being able to transfer the energy from the proximal muscle groups to the more distal ones. Moreover, Pandy & Zajac (1991) suggested that the ankle plantarflexors (soleus, gastrocnemius, and the other plantarflexors) contribute appreciably to trunk power during the final 20% of the jump. This contribution is hard to measure for individual muscles but dynamical simulations have been used to estimate these individual muscle roles and muscle control of the energy flow between segments (Anderson & Pandy, 2003; Delp et al., 2007; Delp et al., 1990; Pandy, 2001; Zajac & Gordon, 1989).

The aim of this study was to use IAA to quantify the contributions of lower limb joint moments to the CM vertical acceleration during a one leg hopping exercise. We believe that this technique will give a better insight on the understanding of the musculoskeletal synergies occurring during this task.

5.3 Materials and Methods

5.3.1 Experimental procedure

Five subjects, three women and two men, participated in this study. They were all physically active, none of them had history of lower extremity injury or pain and their mean age was 25.8±6.6 years. Informed consent was given by each subject prior to testing and the work was approved by the ethics committee of the Faculty of Human Kinetics, Technical

University of Lisbon. Motion capture was collected with a ten camera Qualisys system (model: Oqus-300) operating at 200Hz. Twenty-four reflective markers were placed on predefined anatomical protuberances (acromial, stern, anterior-superior and posterior-superior iliac spines, lateral and medial femoral epicondyles, lateral malleolus of the fibula, medial malleolus of the tibia, 1st and 5th metatarsal heads. Additionally, semi-rigid marker clusters were placed on the lateral aspect of each thigh, shank and foot segments. These were used for the reconstruction of an eight body segment model (trunk, pelvis, right and left thighs, right and left shanks and right and left feet) using Visual 3D software (Visual 3D Basic RT, C-Motion, Inc., Germantown, MD). The ankle was modeled as a universal joint (plantar/dorsi flexion and inversion/eversion), the knee as a revolute joint (flex/ext) and the hip as a spherical joint [flexion/extension, abduction/adduction (abd/add) and internal/external (int/ext) rotation]. The mass properties were based on Dempster's (1955) anthropometric data and the moment of inertia properties were based on Hanavan's study (1964). Ground reaction forces (GRF) and center of pressure position were collected with a Kistler force plate (type: 9865B), synchronized with the motion capture system.

The task consisted of a sequence of unilateral jumps with the dominant lower limb. The dominant leg was determined by asking the subject which was the leg he/she used to kick a ball. To establish a control parameter for the hopping height, a squat jump (SQJ) was performed prior to the hopping task. The minimum height for the hopping jumps was 60% of the maximum height achieved in the SQJ. The subjects performed both the SQJ and the hops on a contact mat which was used as an instantaneous system of visual feedback of jumping height. A monitor, placed in front of the subject, turned green when the jumping height was met, and red otherwise. They were instructed to keep the hands on the waist to minimize arm motion because the head, arms and trunk were modeled as a single segment (Kepple, Siegel, & Stanhope, 1997a). The subjects were asked to perform the hops until we had at least 2 consecutive, consistent and homogeneous jumps, in terms of height. One hopping cycle began with the contact of the right foot on the ground, continued through the suspension phase and ended with the next right foot contact.

Kinematic and kinetic variables were processed with a lowpass 4th order Butterworth filter with a cut off frequency of 6 Hz. Visual 3D was used in order to compute the net joint moments, joint angles and the center of pressure position relative to the contact foot. The net joint moments were resolved into the proximal coordinate system using the cardan sequence X-Y-Z. The data are presented with the hip extension, knee extension and ankle plantarflexon being positive. The curves were plotted and normalized to the stance phase of the hop and the net joint moments were normalized to the subjects' mass.

5.3.2 Induced acceleration analysis

The dynamic equations of motion can be expressed in the following form (Pandy & Berme, 1988):

(Eq.5-1)

where s the joint accelerations matrix, is the inverse inertia matrix (where the segments inertial parameters and center of mass positions are taken into account), is the joint moments matrix, is the Coriolis terms matrix and is the Gravitational terms matrix.

Given equation 5-1, and terms are set to zero allowing us to obtain the accelerations produced by each one of the joint moments (Kepple et al., 1997b). According to Kepple, Siegel, & Stanhope (1997b), it is assumed that the contribution of a joint moment to the acceleration can be determined by applying that moment while considering all other joints to be frictionless and with no torque or stiffness. Thus, because joint stiffness is assumed to be zero, this term does not appear in the equation. The mass matrix in equation 5-2 is a function of the joint angles at any instant in time, thus the magnitude of the computed accelerations is dependent, not only on the moment magnitude ( ), but also on the configuration of the body segments ( ):

(Eq.5-2)

The contribution of each joint moment to angular acceleration was calculated using Visual3D's IAA Module, which solved equation 5-2 for each input net joint moment. Each subject's joint and segment positions were obtained from the motion capture and served as input to the model. Gravity and all net joint moments were set to zero. Each of the joint moments was then entered into the equation and the resultant joint angular accelerations were computed. The input joint moment was then set back to zero, and another joint moment or gravity was sequentially entered into the model. The analysis was done instantaneously, frame by frame, during all the frames of the contact phase of the hop.

5.3.3 Foot-floor interaction

In this study two different contact models were used to establish the connection between the foot and the environment. In the first one, here designated as free-foot system, the foot-floor interface was modeled using a hinge joint with the axis of the hinge passing through the center of pressure in a direction parallel to the medial-lateral axis of the foot. In the second model, called fixed-foot system, the foot was constrained to the floor so that no motion of the foot was permitted.

We firstly chose a free model to explain the joint moments' contributions to the vertical acceleration of the CM but this model didn't seem to explain everything. During the hopping trials analyzed in this study, the heel never contacted the floor and for a brief period of time the foot-floor angle was kept unchanged, presenting an angular velocity close to zero. Thus a fixed-foot system was subsequently applied to each data trial in an effort to parse out the synergies between the muscles crossing the ankle joint and the more proximal lower extremity muscle groups.

The accuracy of the analysis was assessed by computing the absolute mean differences between the vertical ground reaction force obtained experimentally through the force plate and by the sum of all the induced GRF generated by IAA.

5.4 Results

The support phase of the hop cycle was examined by calculating the net joint moments of the right lower limb and by computing the relative contributions of the right ankle, knee and hip joint moments to vertically accelerate the center of mass of the subject (or to the vertical GRF). As illustrated in Table 5-1 (and Fig. 5-1 for one subject, as an example), the accuracy of the analysis was assessed and was below 5,4% of the maximum vertical GRF for all the subjects. This error was assessed by subtracting the GRF data (obtained from the force platform) by the sum of all the induced GRF generated by IAA over the hop stance phase. The absolute mean value corresponds to the absolute mean of all the data points obtained from this difference.

Table 5-1: Absolute mean difference between the experimental GRF and the sum of all the induced GRF generated by IAA over the hop stance phase for the free-foot model, and the correspondent percentage of the maximal GRF obtained with the force plate.

Subject	**Abs mean difference (N)**	**% Max experimental GRF**
1	5.3	0.4
2	6.6	0.5
3	49.7	2.4
4	28.4	1.4
5	76.7	5.4

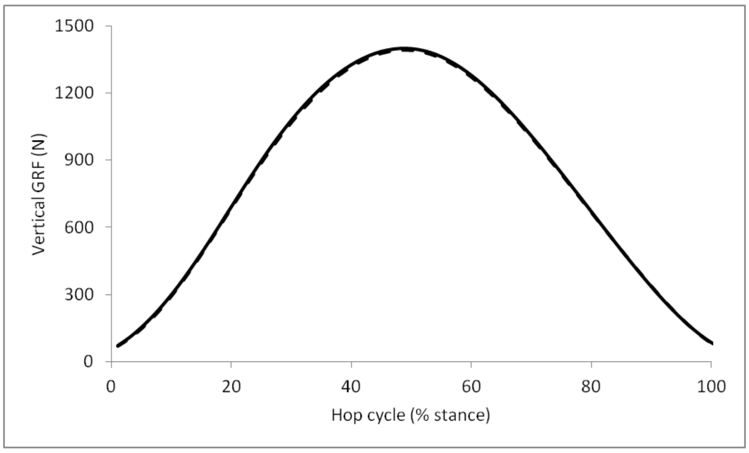

Figure 5-1: Typical subject data of the vertical ground reaction force plot obtained experimentally through the force plate (solid line) and through the sum of all the induced ground reaction forces generated by IAA (dashed line).

5.4.1 Net joint moments of force

As the curve patterns for all the subjects displayed good agreement, the data for one typical subject will be presented in the next figures. Looking at the net joint moments of force (Fig.5-2), all five subjects demonstrated greater moments at the knee, followed by the ankle and the hip.

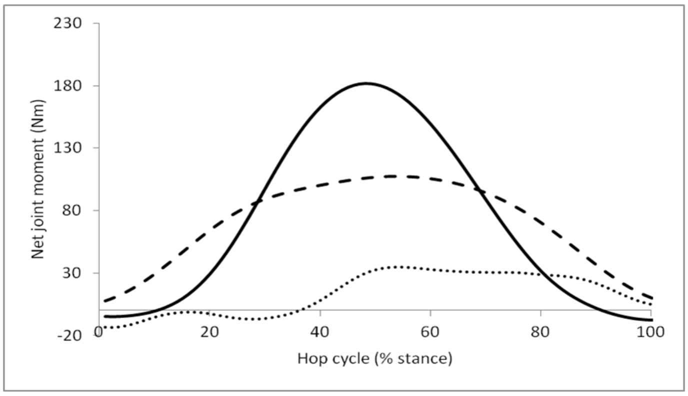

Figure 5-2: Typical subject data of the comparison of the net joint moments of force calculated by inverse dynamics during the contact phase of the hop. Solid line: knee joint; dashed line: ankle joint; dotted line: hip joint

5.4.2 Induced acceleration analysis: free-foot

The vertical acceleration of the center of gravity generated by the joint moments over the stance phase of the hop can be interpreted with the help of Fig. 5-3, since the GRF has a direct relationship with the acceleration of the body's center of mass. When the free-foot model was used, the estimated contributions of the ankle plantar flexor moment to the vertical GRF were found to be much higher than the knee contribution. However, analyzing the induced rotational acceleration of the "joint" between the foot and the floor (Fig.5-4) it is possible to observe that, for all subjects, both the ankle and the knee moments induce nearly equal and opposite acceleration, while almost no contribution of the hip moment is seen. Thus, it is possible that the net action of the ankle moment is to stabilize the foot, with little overall acceleration, and it is likely the two muscles groups are acting in a synergistic way.

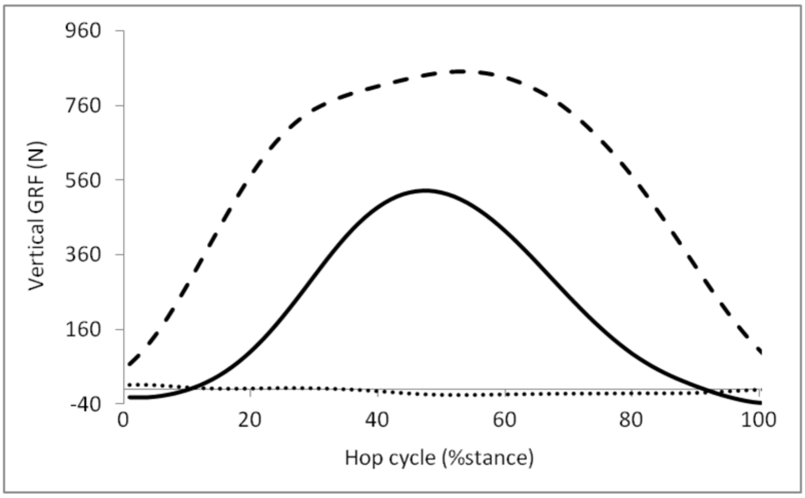

Figure 5-3: Typical subject data of the comparison of the contribution of the ankle (dashed line), knee (solid line) and hip (dotted line) joint moments to the vertical ground reaction force, using the free-foot model.

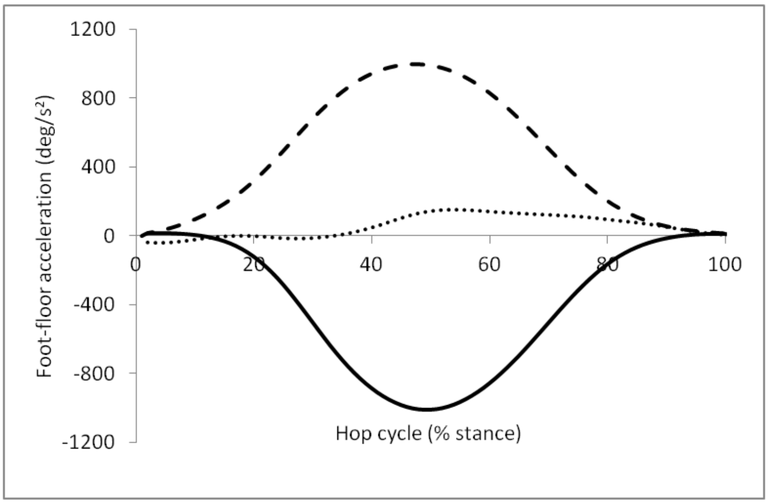

Figure 5-4: Typical subject data of the comparison of the contribution of the ankle (dashed line), knee (solid line) and hip (dotted line) joint moments to the rotational acceleration of the foot-floor “joint”, using the free-foot model.

5.4.3 Induced acceleration analysis: fixed foot

From examination of the ankle angular kinematics, it was observed that there is a period of time where the movement of the foot is small. This time period was defined as the time window where the angle between the foot and the floor is constant during the stance phase (data not shown). Moreover, and looking at the induced acceleration of the foot in Fig. 5-4, it can be seen that the rotational acceleration of the “foot-floor joint” in the free-foot model is close to zero because the contributions of the ankle and the knee are in the same proportion but in opposite directions. Assuming this immobilization period, a fixed foot model was used to investigate the contribution of the ankle, knee and hip, when working with this stable base. In Fig.5-5 it can be seen that the contribution of the knee moment to the vertical GRF is substantially larger than the other moments’ contribution. This was observed in all subjects. With this fixed-foot model, the vertical acceleration of the body would be mostly produced by the knee extensors (which have the larger induced vertical GRF), followed by the ankle moment. In addition, the foot-floor “joint” doesn’t have any acceleration because if the foot is fixed to the floor and has no rotation, it doesn’t make sense to look at this joint. In this particularly case, the contribution of the joint moments is distributed by the other joints, being the knee extensors responsible for the vertical motion of the body.

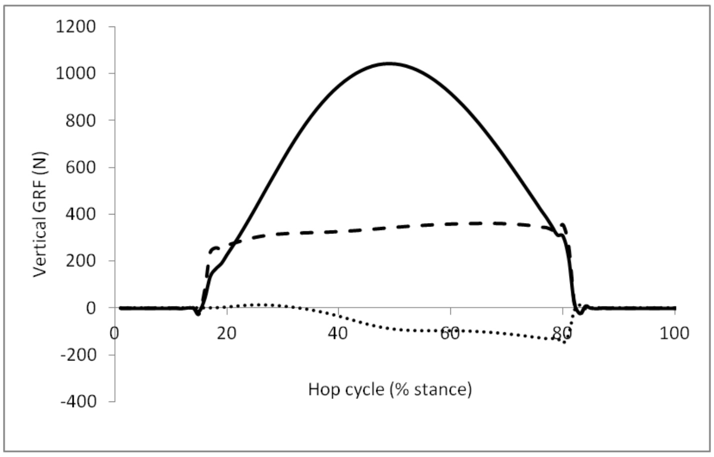

Figure 5-5: Typical subject data of the comparison of the contribution of the ankle (dashed line), knee (solid line) and hip (dotted line) joint moments to the vertical ground reaction force, using the fixed-foot model

5.5 Discussion

We began our induced acceleration analysis using a "free-foot" model containing a one degree of freedom hinge joint connecting the foot and the floor. This model revealed that the action of the knee extensors during the first part of the hop produced a downward rotation of the foot relative to the floor. However, the action of the knee extensors during this period didn't occur in isolation; if the knee extensors were acting alone they would simply drive the heel back to the ground and no upward acceleration would occur. Instead, the downward rotation of the foot generated by the knee extensors was nearly perfectly negated by an upward rotation induced by the plantar flexors. This result likely indicates that there is a plantar flexor moment of force that supplies a stable base from which the proximal knee extensor moment can actuate. In other words, it is likely that the plantar flexors' role is to "turn" the model in a "fixed" foot model, allowing the knee extensors to accelerate the body upward.

Based on the results of the one degree of freedom contact model, we then built a second induced acceleration model in which the foot was fixed to the floor during the time interval when the foot was found to have very small rotational accelerations. This fixed-foot model allowed us to determine the role of the knee and hip net joint moments of force under the assumption that the ankle plantar flexor moment was providing a stable base for the more proximal joint moments to act. It was found that the vertical ground reaction force, which produces support and propulsion, was generated primarily by the knee extensor muscles. To reinforce this statement, another indicator that can be added is the measurement of a sagital plane torque on the force platform. As such, we have calculated the sagittal plane torque of the force plataform (Mx'):

(Eq.5-3)

where Mx is the plate moment about the X-axis and is calculated as:

(Eq.5-4)

where , , and is the vertical force measured by the four sensors of the force plate, is the anterior-posterior force and is the top plane offset. We have calculated the mean and standard deviation of the peak torque from each subject (105±44Nm), thus strengthening the fact that the environment is imposing a constraint such that the use of the fixed model is adequate.

As we get toward the end of the hop, our one degree of freedom contact model revealed that the upward rotation produced by the ankle plantar flexors was much larger than the downward acceleration produced by the knee extensors. Thus, during the later portion of the hop, as the foot rotates forward and the subject leaves the ground, the one degree of freedom model was deemed appropriate. During this interval of foot forward rotation the vertical ground reaction force was found to be primarily due to the action of the ankle plantar flexors.

Induced acceleration can provide an objective decomposition of movement dynamics although results may differ depending on the model used (Chen, 2006). A question arises: if induced acceleration results depend on the formulation of the models, then how can we choose the proper model to explain the dynamics of a task based on IAA results? Or, does an induced acceleration contribution represent a meaningful description of a task function? Despite Chen's critics (Chen, 2006) we believe it does. By presenting results from multiple models, we aimed to show that different models, with different degrees-of-freedom, can be reconciled and used to explain the mechanical behavior during a certain task. It all depends on the aim of the study, the complexity of the task (and consequent decomposition of the movement in different phases) and the motor control theoretical background.

Looking at the free-foot model we found a synergy between ankle and knee muscle groups (Fig.5-4). This result is in line with Bernstein's control theories (Bernstein, 1967) that state that when generating a reproducible pattern of motor behavior, subjects learn to use an algorithm to reduce the number of degrees of freedom. Thus our use of a fixed foot IAA model makes sense during the period of time when the foot is undergoing little rotational acceleration. Based on this model, we believe that the hopping exercise is a synergy between the knee extensors and the ankle plantar flexors with most of the vertical acceleration being produced by the knee extensors (which were found to have smaller induced ground reaction forces when using the free-foot model – Fig.5-2), while the ankle plantar flexors act as stabilizers of the foot (having the larger induced ground reaction forces when using the free-foot model – Fig.5-2). This synergy between the ankle and the knee is perhaps a mechanism that allows the transfer of power from the knee muscles to the ground, and we believe that in this particular task the net action of the foot and ankle moments is to produce a stable foot with little overall acceleration. Moreover, the biarticular nature of the gastrocnemius could be responsible for the transfer of part of the knee extensor moment of force to the plantar flexor moment observed at the ankle joint.

The idea is basically that we should look at the data and have a rationale based in the complexity of the task in order to choose the best set of model constraints. Common optimization techniques assume that the problems of motor redundancy can be solved optimizing cost functions where the solution is unique (Latash et al., 2008; Prilutsky & Gregory, 2000). However, to solve problems of motor redundancy, that unique solution should be replaced by a group or family of solutions, equally victorious in solving the target motor task. The IAA approach, in addition to the joint moments' analysis, may give us increased knowledge and insight regarding muscle groups' function and, therefore, a better understanding of muscle's group coordination. Induced Acceleration analysis is an important augment to the traditional kinematics and inverse dynamic analysis that are commonly used in Biomechanics (especially in gait tasks), but we believe there is a growing need for this technique not just in clinical environment but also in the sports context.

5.6 Conflict of interest statement

The authors have no personal or financial conflicts of interest related to publication of the present work.

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Chapter 6

6 Muscle tendon behavior in single-leg cyclic hopping[4]

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6.1 Abstract

Jumping is an activity that can inflict extreme loads to the musculoskeletal system. Muscles actuate hopping by developing forces that impel the body vertically. To understand this contribution of the lower limb muscles to the vertical acceleration of the body's center of mass during hopping, it was used a muscle-actuated simulation driven by 92 musculotendon actuators of the lower extremity. An induced acceleration analysis was performed and the results present two distinct patterns of muscle contribution. One group demonstrated a similar contribution of the quadriceps and soleus muscles in the total acceleration of the center of mass, in both phases of the stance. The other group demonstrated a time sequence pattern, starting with the soleus muscle contributing with the majority of the acceleration during the braking phase of the support, followed by the quadriceps muscles and ending with the gastrocnemius medialis contribution, which occurs more visibly in the propulsion phase.

6.2 Introduction

Computationally advanced biomechanical techniques have been used to study the role of joint moments and/or individual muscle forces to accelerate the body's center of mass or to accelerate the different joints. Net joint moments, calculated from ground reaction forces, kinematic data and inertial parameters, describe the effect of all muscles that together span a joint (João & Veloso, 2013). Although inverse dynamics are common and well accepted in biomechanical analyses, it is a limited approach, since a net joint moment can be the result of several combinations of muscle forces (Schache et al., 2011)

The forces produced by individual muscles can be estimated using musculoskeletal models and optimization techniques, allowing for the computation of the contribution of each muscle to accelerate the body's center of mass or to accelerate a certain joint (Zajac & Gordon, 1989). These muscle induced acceleration (IA) studies, mostly applied to normal gait (Correa, Crossley, Kim, & Pandy, 2010; Correa & Pandy, 2012; Dorn, Lin, & Pandy, 2012; Goldberg & Kepple, 2009; Liu, Anderson, Pandy, & Delp, 2006; Liu, Anderson, Schwartz, & Delp, 2008; Neptune, Kautz, & Zajac, 2001; Neptune & McGowan, 2011; Pandy & Andriacchi, 2010; Lin, Kim, & Pandy, 2011), impaired gait (Arnold, Anderson, Pandy, & Delp, 2005; Siegel, Kepple, & Stanhope, 2006, 2007), running (Hamner, Seth, & Delp, 2010; Neptune & McGowan, 2011) and pedaling (Neptune, Kautz, & Zajac, 2000) are based in the fact that body segments are kinetic chains of multiarticulated bodies that cause a dynamic coupling effect. Dynamic coupling means that a muscle acting concentrically or eccentrically, produces acceleration on body segments that it does not cross. That is why, for fully understand individual muscle action and to estimate the potential damaging of mechanical loading at the muscle-tendon complex, a modeling approach is necessary to study this causal effect.

These high levels of mechanical demand can inflict extreme loads to the musculoskeletal system and cyclic activities with high magnitude of loads may increase the risk of musculoskeletal injury (McNitt-Gray, 1993). Therefore, the functional role of the lower limb muscles during a given activity needs an important consideration and evaluation.

Regarding jumping and, in particular, hopping, nothing has been reported yet. In this type of tasks which evolve cyclic stretch-shortening cycles, muscle action is known to enhance muscle output in the final of the concentric phase, when compared with the pure concentric action. One advantage of the stretch-shortening cycle muscle action is the ability of the muscle-tendon complex store elastic energy during the eccentric phase, which can be recoiled in the following concentric phase. It became adequate the use of musculoskeletal modeling techniques to add information regarding individual muscle participation in stretch-shortening actions, not just in terms of muscle force production but also in the estimation of the changes that occur in the architectural parameters such as fiber and tendon length, which normally are determined using in vivo methodologies, like ultrasonography (Lichtwark & Wilson, 2005; Mademli, Arampatzis, & Walsh, 2006; Maganaris & Paul, 1999; Magnusson, Narici, Maganaris, & Kjaer, 2008). Vertical jumping is a multiarticular movement characterized by a stretch-shortening cycle with an explosive and elevated power production. These characteristics, associated to training routines, may result in various types of injuries. To improve performance and to reduce injury risk it is relevant to know the load imposed to each muscle and this could be achieved using IA analysis. In sports, possible reasons can be associated to errors during the training programs or derived from the athlete’s technique. Coaches and researchers search for the knowledge of the role of specific muscles during jumping performance, thus knowing how the individual leg muscles contribute to accelerate the body during hopping may lead us to infer about their contribution to the contact forces acting at the hip, knee and ankle joints (Pandy & Andriacchi, 2010).

Therefore, the purpose of our study is to determine which lower limb muscles contribute the most to the vertical acceleration of the body’s center of mass during unilateral hopping. To achieve this, a musculoskeletal model provided by OpenSim software (Delp et al.,

2007) with 92 musculotendon actuators representing 76 muscles of the two lower limbs and torso was used.

6.3 Materials and Methods

6.3.1 Kinematic and Kinetic data

Six subjects, three women and three men participated in this study. They were all physically active, none of them had history of lower extremity injury or pain and their mean age was 25.8±6.6 years. Informed consent was given by each subject prior to testing and the work has been approved by the ethical committee of the Faculty of Human Kinetics, Technical University of Lisbon.

Motion capture was collected with 10 cameras Qualisys (model: Oqus-300) operating at 200Hz. Twenty-four reflective markers were placed in predefined anatomical protuberances (acromial, stern, anterior-superior and posterior-superior iliac spines, lateral and medial femoral epicondyles, lateral malleolus of the fibula, medial malleolus of the tibia, 1st and 5th metatarsal heads. Additionally, semi-rigid marker clusters were placed on the lateral aspect of each thigh, shank and foot segments. These were used for the reconstruction of an eight body segment model (trunk, pelvis, right and left thighs, right and left shanks and right and left feet). The mass properties were based on Dempster's anthropometric data (1955) and the moment of inertia properties were based on Hanavan's study (1964). GRF was collected with a Kistler force plate (type: 9865B), synchronized with the motion capture system.

The task consisted in a sequence of unilateral jumps with the dominant lower limb (hopping). To establish a dynamical control parameter for the hopping height, a squat jump (SQJ) with both legs was performed prior to the hopping task. The minimum height for the hops was set to 60% of the maximum height achieved in the SQJ. The SQJ height was determined by the vertical displacement of the subject's center of mass recorded during the motion capture. The subject performed the hops on a contact mat placed on the top of a force plate, facing a computer monitor where they could see the feedback of their hopping height, calculated from the flight time. They were instructed to keep the hands on the waist to minimize arm motion because the head and trunk were modeled as a single segment and the upper limbs were not included in the model. The subjects were asked to perform the hops until we had at least 2 consecutive, consistent and homogeneous jumps, in terms of height.

6.3.2 Electromyography

The electromyographic activity (EMG) of the tibialis anterior (TA), gastrocnemius medialis (GM), soleus (SOL), vastus lateralis (VL) and long head of the biceps femoris (BF) muscles was recorded. The participant's skin was shaved, gentle abraded and cleaned with alcohol. Afterwards, the surface electrodes (Ambu Blue Sensor N-00-S/25) were placed with an interelectrode distance of 20mm, in accordance with the SENIAM Project recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). A ground electrode was placed over the C7 vertebrae. The EMG data were transmitted by telemetry (Glonner Biotel 88 system) and collected at 1kHz. The EMG signals were processed through a 32bit A/D board synchronized with the force platform and motion capture system. Next, the EMG signals were rectified and low-pass filtered using a bidirectional, fourth order Butterworth filter with a cut-off frequency of 6Hz and normalized to peak activity recorded in the SQJ.

6.3.3 Musculoskeletal modeling

A musculoskeletal model developed in OpenSim software (version 2.4) was used to generate a subject-specific simulation (Delp et al., 2007). Kinematic and ground reaction force data served as input into a 3D model that included 8 segments (two feet–the experimental kinematics only tracked the ankle joint and both metatarsal-phalange and subtalar joint were kept locked -, two tibias, two femurs, one pelvis and one torso with head) and 92 Hill-type musculotendon actuators representing the major lower extremity muscles (Delp et al., 1990; Thelen & Anderson, 2006). In this model, the ankle and the knee were modeled as revolute joints (1 degree of freedom) and the hip and lumbar joints as spherical joints (3 degrees of freedom). The model was manually scaled to match the subject's anthropometry based on his mass and height.

An inverse kinematics technique, based on Lu & O'Connor (1989) global optimization method, was used to minimize the effect of the soft tissue and measurement error. The joint moments of force were calculated by inverse dynamics using a residuals reduction algorithm (RRA) that compensated for dynamic inconsistencies between the measured kinematics and ground reaction forces (Delp et al., 2007). RRA uses the inverse dynamics result calculated from joint kinematics and experimentally measured ground reaction forces, and reduces the magnitude of the residuals by adjusting the joint kinematics and the model mass properties (Hamner et al., 2010). Muscle excitations required to track the kinematics generated in RRA were calculated using a computed muscle control (CMC) algorithm (Thelen & Anderson, 2006; Thelen, Anderson, & Delp, 2003) that distributes force across synergistic muscles. CMC solves a static optimization problem to resolve muscle redundancy by minimizing the sum of the square of muscle activations. Muscle and tendon dynamics accounts for the force-length-velocity properties of muscle and the elastic properties of the tendons (Zajac & Gordon, 1989).

Afterwards, an IA analysis was used to compute the contributions of individual muscles to the acceleration of the body mass center (Hamner et al., 2010; Zajac & Gordon, 1989). The foot-floor interaction model used is well described in Hamner's and colleagues work (2010), being described as a pure rolling contact model where the contact is modeled as a sphere with penetration and friction coefficients simulating the contact of the metatarsal heads to the ground. With this model it was used a static friction coefficient of 0.65 and the contact radius was 0.01m. To perform the IA analysis, the equations of motion for multibody dynamic systems were solved.

(Eq.6-1)

where is a vector of inertial forces and moments; is a vector of centrifugal and Coriolis forces and moments; is a vector of gravitacional forces and moments; is a matrix of muscle moment arms; is a vector of muscle forces; and is a vector of external forces and moments applied to the body by the environment. And this equation, due to the properties of the mass matrix (Pandy & Andriacchi, 2010), can be solved for the accelerations that cause each of the forces as follows:

(Eq.6-2)

In the end it was verified that the sum of the accelerations due to all the forces (muscles, gravity and velocity effects) was equal to the total acceleration of the center of mass of the body calculated through the GRF experimental data, showing the mechanical consistency of the model.

6.4 Results

The purpose of this study was to determine the contribution of lower limb muscles to vertically accelerate the center of mass during a hopping exercise, by analyzing a muscle-actuated simulation during the contact phase of the hop. The accuracy of the simulations was ascertained by comparing the simulated joint angles (RRA), joint moments and muscle activations (CMC) with experimental measured data. The muscle-actuated simulation tracked the joint angles obtained from the experimental data (Fig. 6-1) and we can see that the curves are very similar, with the hip actions presenting some small deviation.

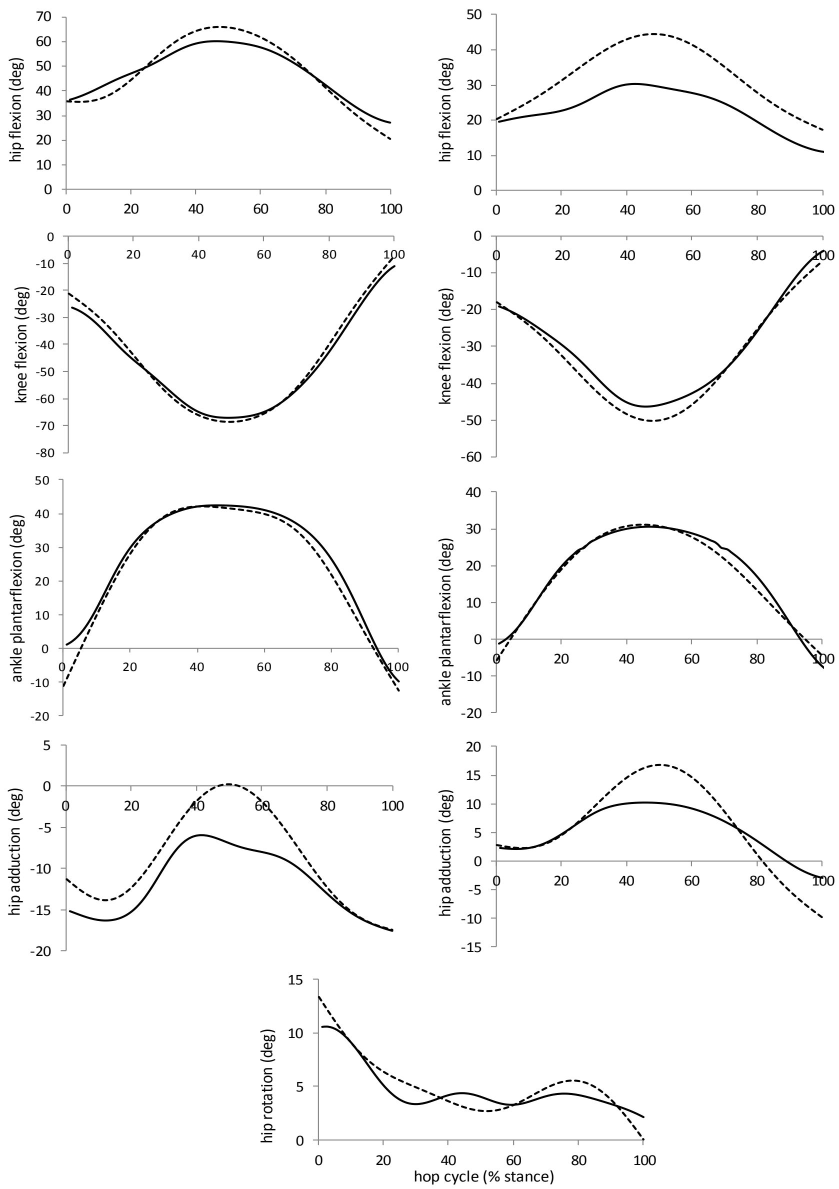


Figure 6-1: Kinematics of the hip, knee and ankle joints during the hopping cycle. Solid line: joint angles calculated by inverse kinematics. Dashed line: joint angles calculated after RRA. Left side: group 1, right side: group 2.

The RMS difference in joint angles was low and the peak residual forces were below 20N. The joint moments calculated by inverse dynamics in RRA module present a matching curve pattern when compared with the joint moments generated by all muscles in CMC (Fig. 6-2), computed as the product of the muscle force and the muscle moment arm. Here we can see a shift to the right on the knee and ankle moments of force.

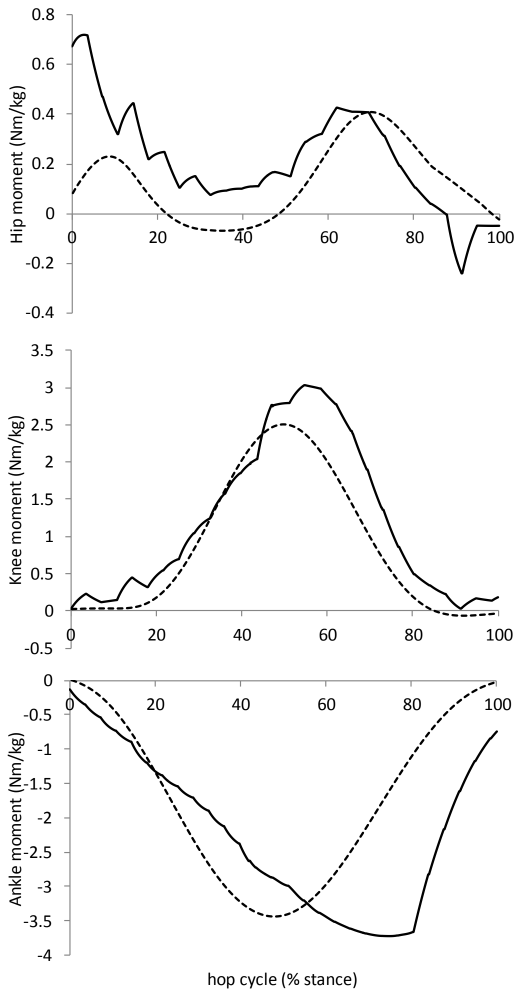


Figure 6-2: Joint moments of the hip, knee and ankle during the hopping cycle, normalized by body mass. Dashed line: joint moments calculated in RRA. Solid line: joint moments obtained by summing the moments generated by muscle forces during the CMC.

Simulated muscle activations were compared with experimental rectified EMG (Fig. 6-3) and the results showed interesting and similar features. Figure 6-3 shows an example from one representative subject. In terms of timing and amplitude, both signals closely match and we can see a strong activation of the VL, SOL and GM. The OpenSim muscle model takes into account the electro-mechanical delay between the electrical neural signal sent to the motorneuron junction and activation, being unnecessary to shift the EMG to enable a better comparison.

The main contributors to the vertical acceleration of the center of mass were selected based on the magnitude of the acceleration (higher than 1m/s^2). From all the muscles of the model, the selected ones were: the quadriceps group (the sum of the three vasti portions and the rectus femoris), soleus, gastrocnemius medialis, gastrocnemius lateralis and peroneus longus. The results during braking and propulsion phases of the hop reveal two interesting and different patterns. In one group (designated as group 1 in Fig. 6-4 - left), the subjects show a similar contribution of the quadriceps (about 50%) and the soleus (about 40%) muscles in the total acceleration of the center of mass. This contribution is equitable in both phases of the stance.

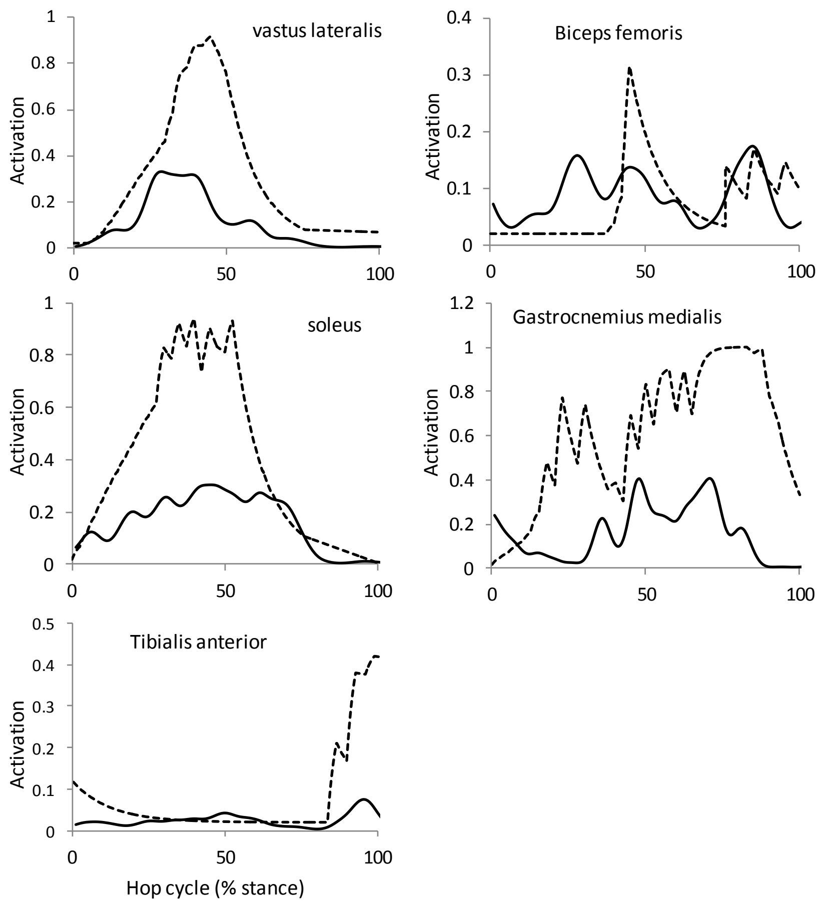
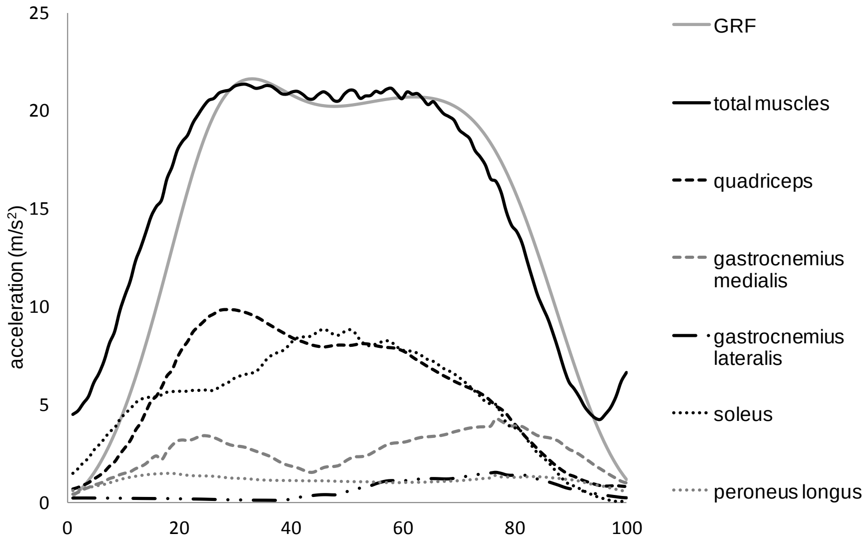


Figure 6-3: Comparison of muscle activations from experimental EMG (solid line) and muscle activations calculated in computed muscle control (dotted line).

Another group (designated as group 2 in Fig. 6-4 - right) presents a time sequence contribution, starting with the soleus muscle contributing with the majority of the acceleration (between 50% and 70% of the total acceleration) during the braking phase of the support, followed by the quadriceps (between 15% and 30%) and ending with the gastrocnemius medialis contribution (between 15% and 25%) which occurs mainly in the propulsion phase. The torso and left limb muscles had a residual contribution (lower than 1 m/s^2).



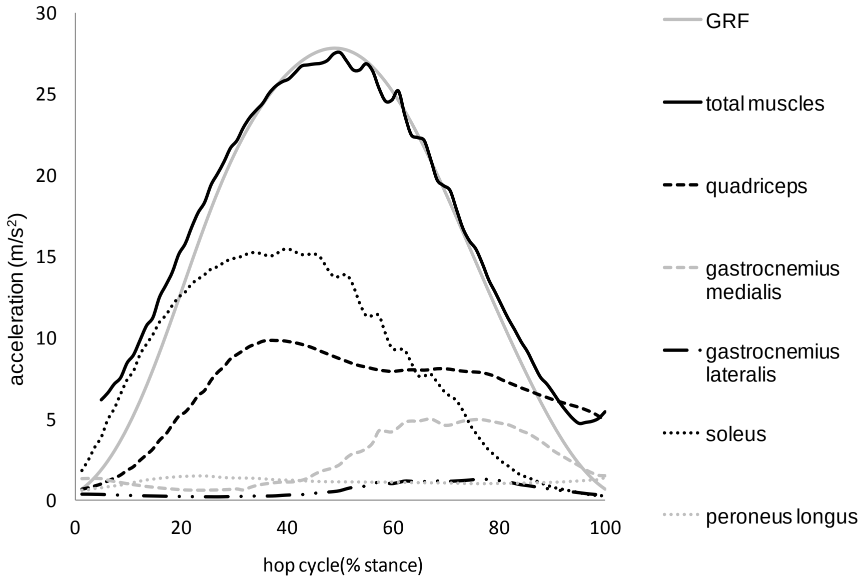


Figure 6-4- Top: Muscle's contribution to vertical acceleration of the body's center of mass, during the contact phase of the hop, for one representative subject from Group 1. Bottom: Muscle's contribution to vertical acceleration of the body's center of mass, during the contact phase of the hop, for one representative subject from Group 2. The solid black line is the sum of contributions of all muscle actuators in the model and the solid grey line corresponds to the acceleration calculated from the experimental GRF.

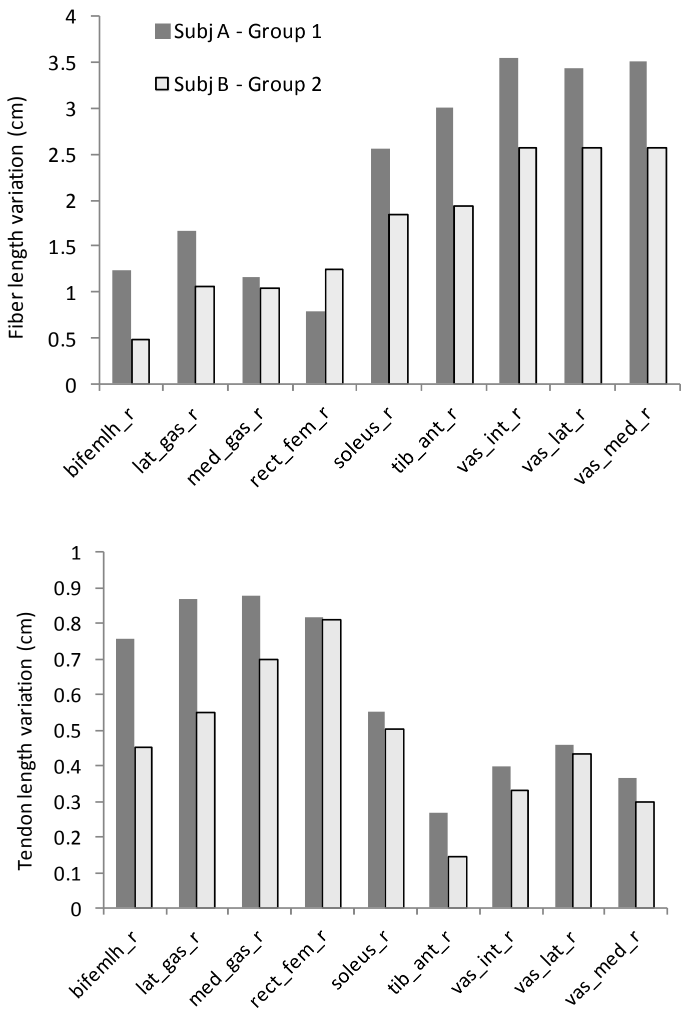


Figure 6-5: Fiber and tendon length variation (cm) during the support phase of the hopping for one representative subject from each group.

Regarding muscle architectural parameters, we can see in Fig. 6-5 the fiber and tendon length variation during the stance phase of the hop for one subject, representative of each group. It can be observed that for the subjects from the two groups, the three portions of the vastus muscle had the highest variation in fiber length (above 2,5 cm) and, correspondently, the less variation in tendon length (below 0,5cm). The tendon structure of the gastrocnemius and rectus femoris muscles achieved the highest variations, reaching 0,9cm in subject A (Fig.6-5).

6.5 Discussion

The purpose of this study was to analyze how lower limb muscles contribute to braking and propulsion during a hopping exercise. This task was accomplished by exploring a three-dimensional muscle actuated simulation of the hopping cycle. It was verified, step by step, that the results of the model nicely agree with the experimental data: 1) the IA results match the ground reaction force data, 2) the muscle excitations are in good agreement with the collected EMG and 3) the estimated morphologic parameters (fiber and tendon length variation) are in accordance with in vivo experiments, concerning muscle and tendon interaction during stretch-shortening cycle actions. The model used in this analysis had over ninety musculotendon actuators, but the vertical acceleration of the body’s center of mass was generated by a relatively small number of muscles. This may be a good indicator that simple muscle coordination strategies are as much as necessary to sustain the vertical acceleration of the body’s center of mass during hopping (Bobbert, van der Krogt, van Doorn, & de Ruiter, 2011; van Soest & Bobbert, 1993). Nevertheless, this more complex model was used also to provide additional insight into the specific muscles involved in the jumping strategy.

After computing the IA results, two groups were defined accordingly with the pattern of muscle’s induced acceleration observed. Group 1 (Fig.6-4 top) demonstrated a similar contribution of the quadriceps and soleus muscles in the total acceleration of the center of mass, in both phases of the stance. Group 2 (Fig.6-4 bottom) demonstrated a different pattern, where a time sequence is present: it starts with the soleus being the major contributor to the acceleration during the braking phase of the support, followed by the predominance of the quadriceps muscles and ending with the gastrocnemius medialis contribution, which occurs more visibly at the propulsion phase. Right before the beginning of the propulsion phase, it is clear the decrease in soleus’ contribution, which finishes before the end of the stance. On the other hand, the quadriceps reaches its maximal acceleration when the propulsion phase starts and keeps accelerating (in lower proportion) until the end of the contact. It is possible to identify here a “transfer” in muscle’s contribution to the acceleration of the center of mass: when soleus starts to decrease its contribution, gastrocnemius medialis appears more evident. Observing the input data for the two groups, the differences in the kinematics of the ankle, knee and hip could explain these results. For subjects from group 1, the kinematic pattern of the hop shows higher amplitude in the joints (Fig.6-1-left), the contact phase is longer (data not shown), the braking and propulsion phases are clearly delimited and the GRF curve presents a subtle two hump pattern. The subjects in group 2 hop with the body much stiffer, show smaller angular range of motion in the ankle, knee and hip joints (Fig.6-1-right) and the GRF curve is a clear parabola. Our results show that different movement patterns are associated with different muscle contributions and muscle patterns.

From previous studies it is clear that the plantarflexors, especially soleus and gastrocnemius, the hip extensors (hamstrings and gluteus maximus) and the knee extensors (rectus femoris and vasti) are the key muscles for generating the propulsion phase of this type of jump (Kim et al., 2013; Pandy & Zajac, 1991). Our results are in agreement, except for the hip muscles, where we didn't find any significant contribution. In this case, our task didn't consist in a maximal vertical jump, instead a hopping exercise, where the hip joint presents smaller range of motion and remains much more extended. Besides that, the lower limb muscles are subject to strong stretching load with a substantial eccentric muscle contraction followed by a powerful concentric contraction. Our results showed that the knee extensors were able to develop a strong vertical push and the ankle plantar flexors are both able to resist the compression produced by the knee extensors and to also develop vertical acceleration of the CG. This action is especially clear at the beginning of the braking phase and at the end of the propulsion phase. From our point of view, a synergist action between knee extensors and plantar flexors allowed the optimal control of vertical acceleration probably with adequate use of the viscoelastic and neuromuscular properties of the lower limb muscles. This is reinforced by the results obtained regarding the architectural parameters, where the tendinous structure of the gastrocnemius and soleus muscles presented the highest changes in length, even with individual variation, agreeing with other studies which refer that the properties of the Achilles tendon can contribute significantly to the total mechanical work of the body during one-legged hopping (Lichtwark & Wilson, 2005). In the same way, the fiber length changes were smaller in those muscles particularly in the vasti muscle group, reaching almost 4cm in some of the subjects. The accuracy of the estimation of muscle forces is determined, in part, by the input values of the musculoskeletal model used and are most susceptible to alterations in tendon and muscle fiber lengths (Pandy & Andriacchi, 2010). If it is assumed that the length of the muscle tendon unit remains unchanged and the tendon experiences a significant length variation, the muscle fibers would benefit from the force-velocity relationship since they could work at a lower shortening velocity (Mademli et al., 2006).

Another aspect that influences the estimation of muscle forces and, therefore, its contribution to accelerate the body segments, is the number of joints that a muscle spans. For instance, the biarticular muscles have been reported as important force transmitters from the hip to the ankle joints, during vertical jumps (Gregoire, Veeger, Huijing, & van Ingen Schenau, 1984). The high velocities of the ankle plantarflexion and knee extension achieved in the end of the push-off result in relative low contraction velocities, enabling them to deliver higher amounts of force. In maximum-height jumps, uniarticular muscles generate the propulsive energy and biarticular leg muscles redistribute that energy and fine-tune the coordination (Zajac, 1993, 2002).

Some limitations of this study must be acknowledged. As mentioned above, one is the fact that muscle IA requires knowledge of individual model parameters related to muscle-force properties, such as muscle fiber length, tendon rest length, maximal isometric force and this model uses parameters obtained from cadaveric specimens. Moreover, it is still needed a more precise measurement of tendon, ligament, muscle, cartilage and bone tissue properties in vivo, to combine with computational modeling in order to describe and explain musculoskeletal behaviors.

Multiarticular movement requires the coordination and different interactions of many structures. Due to this complexity, forward dynamic models have been developed and enriched in order to better analyze human movement data. These simulation models allow us not just to replicate experimental data but also to generate movement trajectories that track a certain hypothesized task (Zajac, 1993). In this case, these models allow us the extraction of “non-measurable” variables to better understand the individual contribution of muscles to accelerate the body vertically during a hopping exercise.

6.6 Acknowledgements

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Chapter 7

7 Assessment of lower leg muscles architecture using diffusion tensor MRI: changes with cyclic stretch-shortening exercise[5]

[5] Submitted as:

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7.1 Abstract

Objectives: to obtain and compare in vivo subject specific muscle architectural parameters that could be used to develop detailed musculoskeletal models.

Design: descriptive and comparative study. Muscle architectural parameters were determined before and after a hopping exercise.

Methods: a Proton Density followed by a Diffusion Tensor Imaging sequence were obtained before and after a cyclic loading unilateral jumping (hopping) exercise repeated until total exhaustion, to compare the following muscle parameters from the lower leg muscles: volume, average fiber length, pennation angle, anatomical cross sectional area and physiological cross sectional area.

Results: Our results show that averaged fiber length increased; averaged pennation angle of SOL decreased and maximal anatomical cross section area of gastrocnemius medialis increased.

Conclusions: MRI scanning allows us not only to obtain relevant muscle properties but also to verify that some of these parameters change with exhaustive jumping exercise. This information could be crucial to the development of musculoskeletal models aimed to studying sports movements characterized by moderated to high mechanical load.

7.2 Introduction

During force production, there are several important muscle architectural parameters to take into account: muscle fiber length, physiological cross sectional area (PCSA) and pennation angle (angle between muscle fibers and the tendon plate) are considered the most important. The anatomical cross sectional area (ACSA) is simply the largest cross-section across the whole muscle while the PCSA is the architectural parameter that is directly proportional to the maximum force generated by the muscle, because it takes into account the pennation angle (Kawakami, Ichinose, & Fukunaga, 1998). It is also the parameter most difficult to measure because its value is based on the entire muscle volume and includes the estimate of fiber length to arrive at the value for area across the entire muscle (Lieber, 2002), so it is indirectly determined from the muscle volume and muscle fiber length.

Computational models of the musculoskeletal system are widely used to study the mechanisms of musculoskeletal performance and to simulate specific conditions. They have been created based on anatomical data collected from cadaveric specimens. Models allow us to answer "what if" questions, to isolate individual sources of input data and estimate muscle parameters, such as muscle forces, that are difficult to measure experimentally. To estimate muscle volume or PCSA, data of muscle force-generating capacities derived from cadaveric studies are often used to scale models of the muscle-tendon unit (Anderson & Pandy, 1999; Delp et al., 1990). The same happens with pennation angle: simulation models commonly assume muscle fibers as one single structure with only one angle between the fibers direction and the aponeurosis. Thus, the need to create accurate, individualized models of the musculoskeletal system is driving advances in imaging techniques, especially in magnetic resonance imaging (MRI).

MRI is considered the most useful non-invasive imaging technique that allows the reconstruction of the whole muscle, enabling the quantification of detailed structure, function and metabolism from the muscle tissue. One of the MRI methods is Diffusion tensor imaging (DTI). This imaging technique relies on the correspondence between cell geometry and the anisotropic nature of water diffusion in the muscle. The theoretical basis for DTI states that self-diffusion of water in tissue is restricted by membranes and other cellular constituents, resulting in an apparent diffusion coefficient, which is lower than the free diffusion coefficient and is orientation-dependent for elongated structures. Skeletal muscles are composed of fibers that have an elongated, generally cylindrical shape. The diffusion tensor can be mathematically described by a 3x3 matrix, and at least six independent directions for the diffusion gradients must be assessed to calculate the diffusion tensor. The signal obtained on Proton density (PD) weighted scans consists in a spin echo sequence, with short echo time and long repetition time. PD images are useful for determining the muscles boundaries, due to the elevated detail in the image and tissue differentiation capability, enabling the determination of muscle cross sectional area, and subsequently muscle volume using simple image processing software.

Muscle volume is itself essential when evaluating muscle performance and functional consequences of changes in muscle size and strength due to training, aging or immobilization (Albracht, Arampatzis, & Baltzopoulos, 2008; Kawakami, & Lieber, 2000). These studies reported increases in muscle thickness and pennation angles of vastus lateralis muscle following leg press exercise (Csapo, Alegre, & Baron, 2011), a cycloergometer test until exhaustion (Brancaccio, Limongelli, D'Aponte, Narici, & Maffulli, 2008) or following strength training programs (Blazevich, Cannavan, Coleman, & Horne, 2007). Regarding acute effects of exercise, analyzed with MRI techniques, the following studies have been reported: ankle plantarflexion exercises in the supine position with eccentric contractions showed changes in gastrocnemius medialis (GM) values of functional anisotropy (FA), eigenvalues and apparent diffusion coefficient (ADC) and T2 after exercise (Yanagisawa, Kurihara, Kobayashi, & Fukubayashi, 2011); during a walking exercise, with fully extended knees and ankles dorsiflexed, followed by a 30-s dorsiflexion exercise, the ADC and T2 values of the tibialis anterior muscle (TA) increased (Yanagisawa, Shimao, Maruyama, & Nielsen, 2009); another study showed that after a one-legged plantarflexion exercise, inactive muscles decreased CSA and T2, with no changes in ADC; and the exercised-leg increased CSA, T2 and ADC (Nygren & Kaijser, 2002).

Regarding these results and the lack of information concerning specific muscle architecture parameters after exercise, the aim of our study was to determine the in vivo 3D architectural parameters (volume, ACSA, pennation angle, fiber length and PCSA estimation) of the lower leg muscles, using PD and DTI imaging data before and after an exhaustive unilateral jumping task.

7.3 Methods

Six young female (age: 25.7 ± 5.2 years, mass: 61.0 ± 8.1 kg, height: 1.6 ± 0.1 m), healthy volunteers participated in this study. They were all physically active and none of them had history of lower extremity injury or pain. Informed consent was given by each subject prior to testing and the work has been approved by the ethical committee of the Faculty of Human Kinetics, Technical University of Lisbon.

All the participants were examined in a 1.5T whole-body MRI scanner (Signa HDxT 1.5T, GE Healthcare, USA) before and immediately after the exercise protocol. A pelvic coil was used for all the measurements. This coil was used instead of an extremity coil because we wanted to scan the entire lower leg, from the femoral condyles until the calcaneous, so we could have the entire muscle length. Leg immobilization was carried out by using a custom made structure, made of a nonmagnetic material and adjustable to the subject's leg length, which avoided compression of the leg muscles against the MRI table. This structure was fixed to the table and the participants kept their foot and leg in the same position, guarantying the same reference frame in all the MRI acquisitions (before and after the exercise protocol) and a constant ankle position. The following imaging protocol was performed: first, Proton Density and Diffusion Tensor images were acquired from the dominant lower leg at rest. The images were acquired using the following parameters: for DTI, 54 axial acquisitions, Gradient-Echo sequence, TE=35, TR=3622, slice thickness=3.9mm; and for Proton Density: 54 axial acquisitions with Turbo Spin-Eco Gradient, TE=7.6, TR=4140, slice thickness=3.9mm.

After the first bout of MRI scans, the participants were asked to perform a sequence of one-legged jumps, keeping the hands on the waist. They were constantly receiving verbal encouragement to keep performing the exercise, despite the fatigue increasing. The end of the task was considered when the subject couldn't take the entire foot from the floor anymore (the total exercise time was 1m 30s in average). Immediately after the exercise, the subject was placed again in the MRI table, with the exercised leg positioned on the top of the leg fixation structure and the same bout of imaging sequences was collected.

Regions of interest (ROIs) were drawn in the PD images using a piecewise linear boundary provided by the Osirix software (Rosset, Spadola, & Ratib, 2004) (Figure 7-1-left). The ROIS were outlined manually according with the anatomical boundaries of the Tibialis Anterior (TA), Gastrocnemius Medialis (GM), Gastrocnemius Lateralis (GL) and Soleus (SOL) muscles. All the ROIs were manually drawn by the same operator, with a total of 46-82 contour curves for each muscle on each phase of the protocol (before and after the exercise). The volumes of individual muscles (Vm) were determined by summing the product of the cross-sectional areas of the muscle from each image and the slice thickness (3,9mm). Also from the PD images, the ACSAmax was determined as the largest cross-section outlined across the whole muscle. Muscle length (Lm) was defined as the distance between the muscle's proximal and distal end.

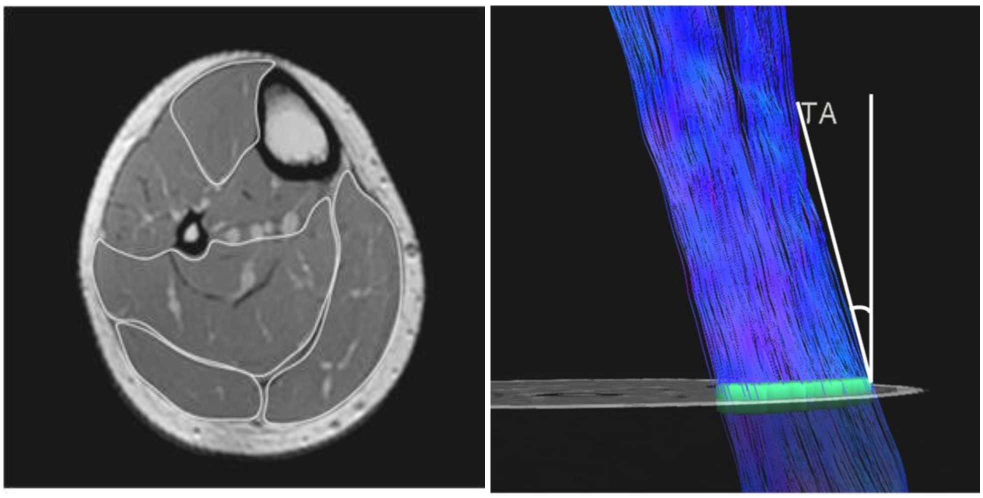

Figure 7-1 Left: Boundaries of the GM, GL, SOL and TA muscles, delimited in the DP images. Right: calculation of the pennation angle for the TA muscle.

The computation of the diffusion tensor imaging results was performed trough a custom script that uses FSL Tools (FMRIB Analysis Group, Oxford, UK) developed to process the imaging data. This routine included the correction of motion and "eddy current" artifacts and the co-registration of the PD images with the DTI images.

Muscle fibers were tracked from the DTI using the free software Diffusion Toolkit and TrackVis (Department of Radiology, Massachusetts General Hospital, USA). The slice correspondent to the ACSAmax of each muscle was chosen to draw the seed ROI necessary for the tracking. Two more ROIs were draw, four slices above and four slices below, to guarantee the tracking of fibers only from that specific muscle. From the tracked fibers, the following parameters were determined: average fiber length (Lf), defined as the mean value of the length of the fibers calculated through the tracking algorithm in Diffusion Toolkit; pennation angle (θ), defined as the fibers angle relative to the force-generating axis. Pennation angle was measured directly in the fiber tracked images using Osirix software (Rosset et al., 2004), by rotating the image in the longitudinal axis and positioning it in a plane where the preferred angular orientation of the fibers was visualized (Figure 7-1-right). When two or more fibers orientations were visualized, all the angles were determined and the mean value was used in the results. PCSA represents the sum of the cross-sectional areas of all the muscle fibers within the muscle. It was estimated using Equation 1, derived from the work of the pioneer Gans & Bock (1965) and continued nowadays by Lieber & Friden (2000).

Eq.7-1

The normality of the variables was tested and a paired-samples t-test analysis was used to compare the differences between the architectural parameters before and after the exercise task. The statistical significance was set at the level of $p<0.05$. All the statistical tests were performed using IBM SPSS Statistics 20 software.

7.4 Results

Regarding the results, Table 7-1 shows the architectural parameters obtained for the GM, GL, SOL and TA muscles in the two conditions: pre and post exercise.

Table 7-1- muscle architectural parameters

	V_m (cm³)		$ACSA_{max}$ (cm²)		L_m (cm)	L_f (cm)		PCSA (cm²)		Pennation angle (⁰)	
	Pre	Post	Pre	Post		Pre	Post	Pre	Post	Pre	Post
GM	189.4±56.3	193.6±54.8	14.0±4.0	14.4±4.1*	23.1±1.8	6.3±0.5	7.0±0.3*	28.3±7.9	26.6±8.2	18.1±2.1	17.9±2.0
GL	96.1±28.5	100.2±29.7	8.3±1.8	8.8±2.4	20.3±1.7	5.8±1.6	7.2±1.0*	16.9±5.8	13.7±4.1	11.3±2.2	13.7±3.1
SOL	369.2±66.5	370.2±62.9	24.9±4.2	25.0±4.1	29.9±1.2	5.6±0.9	6.1±0.7*	59.7±5.6	56.4±4.3	25.7±1.3	20.4±3.0*
TA	110.9±22.8	109.3±21.4	8.1±1.4	7.9±1.2	27.8±2.4	9.3±0.5	9.5±0.9	9.5±2.1	9.3±1.4	19.7±2.9	16.8±2.7

Mean values ± SD of the muscle architectural parameters of the GM, GL, SOL and TA muscles determined by imaging techniques PD and DTI, pre- and post-exercise. The length of the muscle is the same in the two conditions. * significant different (p-value<0.05)

Both muscle volume and $ACSA_{max}$ didn't registered statistically significant differences after the exercise, with the exception of the GM $ACSA_{max}$, which increased significantly in the post-exercise condition. The results show that the SOL was the muscle with higher volume, followed by the GM, the TA and the GL. The $ACSA_{max}$ was higher for the SOL, GM, and GL and lower for the TA. In what concerns the total length of the muscle, the longer muscle was the SOL reaching almost 30cm, followed by the TA, GM and GL muscles.

Regarding pennation angle, it was verified that different portions of the muscle fibers had different pennation angles, especially for the SOL muscle. The preferential directions were assessed although only the mean of those angles is reported. Thus only one pennation angle per muscle was assumed. The highest pennation angle was found in the SOL muscle, followed by TA, GM and GL muscles. After the exercise, only the pennation angle of the SOL muscle revealed a significant decrease.

The fiber length was assessed with the fiber tracking process and the mean length of all the muscle fibers is reported in Table 7-1. The longest fibers were observed in the TA muscle, followed by the SOL, GL and GM muscles. With the exercise, the GM, GL and SOL muscles showed significant increases in the fiber length. These results lead us to a discussion concerning the fiber tracking method itself, which is better explored in the Discussion section.

PCSA was estimated using Eq.7-1, with the measured data V_m, L_f and pennation angle. We can observe that the SOL muscle had the highest PCSA, followed by the GM, GL and TA. After the exercise, no significant changes were observed.

7.5 Discussion

The aim of this study was to determine the architectural parameters of the human lower leg muscles (volume, ACSAmax, muscle length, pennation angle, PCSA and fiber length) using diffusion-tensor and anatomical MRI data, and whether short-term exhaustive stretch-shortening cycle exercise would induce changes in those parameters. To accomplish that, the process started with visualization and delimitation of ROIs corresponding to each muscle in each slice of a DP image scan. Afterwards, the maximal ACSA and the length of the muscle was assessed, followed by a fiber tracking algorithm that mapped the diffusion tensor, allowing for the reconstruction of the muscle fibers, and subsequent estimation of fiber length, pennation angles and PCSA.

Muscle volume didn't change significantly after the acute exercise, although the increased water content in exercising muscles is known to affect both extra and intra-cellular volumes (Fisher, Meyer, Adams, Foley, & Potchen, 1990). One of the reasons for this unchanged result may be the time between the end of the exercise and the duration of the scan protocol. Due to the rapid decay of the DTI signal, the DP scan was performed only after the DTI scan. It is possible that the recovery of muscles has been rapidly done and when drawing the muscles' boundaries to calculate muscle volume, the anatomical images didn't reveal significant differences. Muscle models generally assume that pennation angle is constant across all fibers. In accordance with other imaging studies (Heemskerk, Sinha, Wilson, Ding, & Damon, 2009; Sinha, Sinha, & Edgerton, 2006), we verified that fibers from the same muscle have different pennations. By creating a tridimensional representation of muscle fibers, we can add information that may help us to surpass the limitation of representing muscles as single line segments that remain unchanged in terms of shape when interacting with the underlying bones and other structures (Blemker, Asakawa, Gold, & Delp, 2007). After exercise, the pennation angle significantly decreased in the SOL muscle. This change may be attributed to the challenging determination of pennation angles in this particular muscle due to its multipennated architecture. It is quite difficult to determine all the preferential directions of the tracts and of course the mean value doesn't show exactly the individual fiber pennations. It was expected an increase in the pennation angles of the most contributing muscles (plantarflexors) but no significant changes were verified.

The tractography process enables the quantification of the length of the muscle fibers. Fiber length is a muscle parameter that is commonly estimated through ultrasonographic equipment (Aggeloussis, Giannakou, Albracht, & Arampatzis, 2010; Arampatzis et al., 2006; Bojsen-Moller, Hansen, Aagaard, Kjaer, & Magnusson, 2003; Brancaccio et al., 2008). Ultrasonography and reconstruction methods have been widely used to determine not just fiber length but also pennation angles but they do not take into account spatial variation and orientation of the muscle fibers and can only be used in superficial muscles. Moreover, when dealing with long muscle fibers, the length is not directly measured but instead it is estimated by regression. Another advantage of the DTI is the ability to measure both fiber length and pennation angle through the mapping of the eigenvalues (λ_1, λ_2 and λ_3) from the diffusion tensor, which gives us the direction of the fibers. Having a subject-specific measure of fiber length, and using Eq.7-1, it is possible to estimate the PCSA, thus predicting the ability of that particular muscle to produce maximum force. We observed that the length of the fibers estimated through DTI were higher than the ones reported in literature with cadavers or live subjects' experiments. After the exercise task, the length of the fibers significantly increased for the plantarflexors SOL, GM and GL. We believe that the increase of the water content that circulates inside the cells will facilitate the water diffusion. Increasing the space between the microstructures may have as a consequence the increased diffusion in all the directions of the tissue. This less restriction to the diffusion in the longitudinal axis of the fibers may justify the increase on the "apparent" length of the fibers. The "apparent" length of the fibers before the exercise may have been underestimated since the probability of the tracking to be discontinued is much higher. Moreover, different stop criteria settings change the resulting fiber length. We used a default curvature stopping factor of > 20˚.

Comparing with the existing literature data from cadaveric studies (Wickiewicz, Roy, Powell, & Edgerton, 1983), our results for the pennation angle showed higher values for TA (almost twice the pennation) and similar values for the other muscles. Comparing with in vivo studies using ultrasonography, we found similar results for the anterior angle of pennation of the MG muscle, also for the posterior angle of pennation for the GL muscle and for both compartments analyzed for the posterior soleus muscle reported by Martin and colleagues (2001). Regarding muscle length, our values were smaller for the four muscles (less than 1cm) when compared with the same cadaveric database and also with the subjects assessed by Albracht et al (2008) through ultrasonography. However, our sample was a group of young females, with different anthropometric characteristics and smaller segments. The PCSA estimated revealed much smaller values when compared with the existing literature (although SOL and TA PCSA are similar to those assessed in cadavers). We can attribute that to the estimated fiber length, which may have been overestimated (both in –pre and –post exercise). Moreover, the rigor mortis can produce a slow contraction in the skeletal muscle (Bendall, 1951), with consequent effects in the fiber architecture.

The relationship between structure and function of muscles is critical to understand the physiological basis of force production and movement, especially in adaptive conditions such as training, detraining, disease, age, immobilization, etc, where the architecture of the muscle can undergo changes (Blazevich, 2006; Nimphius, McGuigan, & Newton, 2012). This relationship is usually studied through musculoskeletal models, where the anatomical and architectural parameters of the muscles are usually simplified and/or adapted from cadaveric databases, most of them with a small sample size (Wickiewicz et al., 1983). Structural and functional parameters provided by PD and DTI , such as the trace of the diffusion tensor, allows for the non-invasive and in-vivo measurement of critical parameters for the elaboration of those models, such as muscle volume, fiber length and PCSA.

7.6 Conclusion

This study indicates that DTI is a viable non-invasive tool for assessing muscle architecture, specifically muscle volume, fiber length, PCSA and pennation angle. Muscle architecture has a considerably influence on the mechanical properties of the muscle-tendon complex, thus being crucial in force production. The use of imaging techniques, such as DTI is an important advance in the study of human muscle architecture, promoting the integration of subject-specific parameters into musculoskeletal models, as a replacement for cadaveric-based data. Regarding the use of DTI immediately after exercise, careful should be taken because by increasing the acquisition time, the effect of the exercise may be partially lost during the scan.

7.7 Practical Implications

- Diffusion Tensor imaging allows us to obtain relevant muscle properties and also to verify that some of these parameters change with exhaustive jumping exercise.

- This information is crucial to the development of musculoskeletal models aimed to studying sports movements characterized by moderated to high mechanical load.
- The combination of advanced imaging techniques with computation methods will allow us for the creation of individualized models, and subsequently estimation of muscle forces during specific movement patterns.

7.8 Acknowledgments

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Chapter 8

8 General Discussion and Conclusion

8.1 General Discussion

In the course of this dissertation, our underlying hypothesis stated that the amount of loading can influence the biological tissue morphology and mechanical properties. In particular, high demanding exercise until fatigue can be associated to acute musculoskeletal transformations and compensations. To accomplish this goal, two different but complementary approaches were used: 1) biomechanical modeling techniques such as joint moment of force and muscle induced acceleration and 2) in vivo imaging techniques such as diffusion tensor MRI. Scientifically, the contribution of this thesis is: 1) to increase the knowledge and insight about the use of an induced acceleration approach to quantify the contribution of joint moments and /or individual muscle moments of force to accelerate the body and how that contribution changes with repeated exercise and 2) to use MRI techniques to obtain in-vivo subject-specific data from the architectural parameters of the muscle, in order to refine the musculoskeletal models needed for the estimation of the mechanical load applied to the muscle and joint structures during exercise.

The chapters of this dissertation (from the third to the seventh chapters) correspond to a number of original research articles whose primary goal was to show how different methodologies can be used and combined to provide important information regarding the effect of mechanical loading in the lower limb musculoskeletal system. This would allow us to shed some ideas regarding the use of a multidisciplinary approach when dealing with musculoskeletal modeling. In view of that, this final chapter serves primarily to make an overview of the main findings of the preceding chapters and ends with the next step and some suggestions for future research.

8.2 Overview of main findings

In **Chapter 3**, we investigated the effects of repetitive SSC actions in the kinematics, kinetics and neuromuscular behavior of the lower limb using the hopping model. We corroborate many of the results obtained by other investigators, namely the joint kinematics, where the increased knee flexion and ankle dorsiflexion occur with fatigue, in an attempt to help absorbing the force during impact (Self & Paine, 2001). On the kinetics side, the reduction in GRF could be an indicator of less tolerance to the repeated SSC (Komi, 2000), reducing the efficiency of the muscle-tendon complex to absorb energy. Moreover, the ankle and knee joint power peaks decreased with fatigue and a possible compensation mechanism may have occurred, with an augmented hip power in the final hop. Using the unilateral hopping model it is possible to study the fatigue phenomena and observe the adaptations in the kinematics, kinetics and neuromuscular pattern of the lower extremity. The joint power data clearly supports the loss of power in the ankle and knee joints. These results are in agreement with the neuromuscular activity, which reflected a reduction in the EMG bursts of the plantar flexors (expected for an exhaustive SSC). At the same time, the BF muscle increases its activation, probably to reduce the load in the knee extensors (Kulas, Hortobagyi, & Devita, 2010). The EMG activity has been reported as a subject-specific neuromuscular response, where the inter-individual differences can be assigned to biological and structural differences in muscle tissue, thus affecting its mechanics (Kuitunen, Avela, Kyrolainen, Nicol, & Komi, 2002; Nicol, Komi, Horita, Kyrolainen, & Takala, 1996).

Following this traditional biomechanical analysis of the kinematics, kinetics and neuromuscular behavior of the hopping, in **Chapter 4** we introduced an intermediate stage of interpretative analysis, the induced acceleration analysis, to combine with the descriptive analysis that predominated in Chapter 3. Here we aimed to quantify the contribution of individual net joint moments of force to accelerate the center of mass of the body. This approach has been mentioned with a clinical relevance when dealing with patients with loss of muscle function who will use other muscle groups to accelerate the joints they do not cross (Zajac & Gordon, 1989). Nevertheless, this analysis should also be addressed to fatiguing mechanical tasks such as hopping until exhaustion to understand the strategy used by the muscle-skeletal system to avoid mechanical failure. With this analysis we observed that the relative contribution of ankle plantar flexor moment and knee extensor moment decreased while the hip extensor moment increased with fatigue. The hip moment contribution to the vertical GRF was also found to be negative, which can be attributed to the flexion of the hip that occurred simultaneously with the plantar flexion of the ankle, especially in the final hops where the mechanical performance changed and the force dropped after impact, probably related with the loss of tolerance to the repeated stretch load as fatigue progresses (Komi, 2000). The main advantage of this approach is that this interpretative process allows the data to be presented to the clinician or to the coach in a more comprehensible way, particularly the use of adaptive strategies that rely on the ability of different muscles or muscle groups to accelerate the joint which they not span. (Hamner, Seth, & Delp, 2010; Liu, Anderson, Pandy, & Delp, 2006; Liu, Anderson, Schwartz, & Delp, 2008).

The induced acceleration analysis has been criticized by some investigators due to the different results obtained, depending on the model used (Chen, 2006). To gain more insight concerning this issue, in **Chapter 5** we tested whether the use of two different foot-floor models would reveal different induced acceleration results and if so, what would be the consequences. With a free-foot model (a one degree of freedom hinge joint connecting the foot to the floor) the knee extensors downward rotated the foot while the ankle plantar flexors perfectly negated that movement producing an upward rotation of the foot. The fixed-foot model assumed this stable base provided by the plantar flexors and enabled us to determine how much the proximal joints would contribute to accelerate the body upwards. With this model, the knee extensor moment produced most of the vertical GRF. These two models demonstrated that synergies occur and can be interpreted according with the complexity of the task, in order to choose the best set of model constraints. If subjects learn to use the minimum number of degrees of freedom (Bernstein, 1967), it makes sense for us to use a fixed-foot model during the floor contact where no rotational acceleration occurs and a free-foot model during the late period of the propulsion phase, when the foot needs to get off the ground. This interpretative analysis of the contribution of the joints to accelerate the body leads us to a better comprehension of how muscles work together for the same purpose yet not equally. However, this analysis doesn't permit to know which individual muscle does what. In an attempt to answer this question, **Chapter 6** presents a study where an induced acceleration was also performed but having as input not the joint moments of force but instead each individual muscle force from a model comprising 92 actuators representing the muscle-tendon structures of the lower limbs and trunk. We observed that in hopping the vertical acceleration was generated by a small number of muscles, indicating that simple muscle coordination strategies are as much as necessary to move the body (Bobbert, van der Krogt, van Doorn, & de Ruiter, 2011; van Soest & Bobbert, 1993). Two different patterns were observed: in subjects with an increased joint amplitude, both quadriceps and soleus muscles contributed similarly for the total acceleration of the body; in subjects with a stiffer hopping pattern, the soleus muscle contributes with the majority of the acceleration during the braking phase of the support, followed by the quadriceps muscles and ending with the gastrocnemius medialis contribution, which occurs more visibly in the propulsion phase. The knee extensors develop a strong vertical push and the ankle plantar flexors are both able to resist the compression produced by the knee extensors and to also develop vertical acceleration of the center of mass. The tendinous structure of the gastrocnemius and soleus muscles presented the highest changes in length, agreeing with other studies which refer that the properties of the Achilles tendon can contribute significantly to the total mechanical work of the body during one-legged hopping (Lichtwark & Wilson, 2005). Assuming that the length of the muscle tendon unit remains unchanged and the tendon experiences a significant length variation, the muscle fibers would benefit from the force-velocity relationship since they could work at a lower shortening velocity (Mademli, Arampatzis, & Walsh, 2006). One limitation of using muscle-skeletal models with muscle-tendon actuators is the fact that individual model parameters related to muscle-force properties, such as muscle fiber length, tendon rest length or maximal isometric force are obtained from databases with a small number of cadaveric specimens. The accuracy of the estimation of muscle forces is determined, in part, by the input values of the musculoskeletal model used and are most susceptible to alterations in tendon and muscle fiber lengths (Pandy & Andriacchi, 2010). In order to surpass this issue, an initial approach is presented in **Chapter 7**, the final study of this dissertation, where we measured in-vivo the muscle architectural parameters of the lower leg that influence the most the ability of the muscle to produce force (volume, anatomical and physiological cross section area, muscle length, pennation angle and fiber length). Data provided by diffusion tensor imaging and tractographic processes enabled the reconstruction of the muscle fibers, and subsequent estimation of its length, pennation angles and physiological cross section areas. By creating a tridimensional representation of muscle fibers, we can add information that may help us to surpass the limitation of representing muscles as single line segments that remain unchanged when interacting with the underlying bones and other structures (Blemker, Asakawa, Gold, & Delp, 2007). The quantification of muscle volume, by delimiting its individual area on each slice of the imaging data and across the length of the muscle, gives us an architectural parameter needed to calculate subjects-specific physiological cross section area. The pennation angle was a challenging parameter to obtain because the tridimensional fiber tracts show a great variability in the pennation of the muscles, especially soleus. The preferential direction of the fibers is ample and a mean value of the pennation angle was calculated and used for comparison between the pre- and post-exercise condition. This is limitative but for this particular purpose we chose to calculate the mean angle of pennation. The increase of the water content that circulates inside the cells will facilitate the water diffusion, which may have as a consequence the increased diffusion in all the directions of the tissue. The length of the fibers estimated by the tractography is dependent on this degree of restriction and also on the stopping factor which was manually set at 20^{o}. The information collected in this study will be used to refine musculoskeletal models in order to better predict the individual muscle force and joint reaction forces of a particular subject. The combination of advanced imaging techniques with computation methods will allow us to individualize models, and subsequently estimate muscle forces during specific movement patterns. Moreover, subject-specific in-vivo data can be used instead of cadaveric data.

8.3 Future research

Given the present dissertation, the limitations presented on each study and the work developed so far, I believe that the next steps will definitely be the sequel of the work elaborated with the diffusion tensor imaging techniques and the muscle modeling approach. The use of the diffusion tensor to quantify the tridimensional muscle architecture and also to locate the origin and insertion points of each muscle give us subject-specific information that must be used in the elaboration of accurate muscle-skeletal models. Of course in this case we only have data from the lower leg of the subjects but it is a starting point. Muscle-skeletal models need to have subject-specific and in-vivo parameters from the maximal isometric force that can be produced by each muscle, pennation angles that best represent each compartment of a multi-pennated muscle and accurate fiber and tendon lengths. Changing these parameters in the standard models and simulating the motion afterwards can give us a better insight regarding the factors that affect the production of force.

In addition, for my next goals, I would like to use both muscle architectural data (obtained with imaging techniques) and muscle-skeletal modeling applicable in the clinical context, in particular regarding the injury during sports performance and pathologies that affect the locomotor system. This is a field of investigation that has become tremendously relevant in modern society and our laboratories and clinical facilities are now getting more aware and conscious to the benefits of including a biomechanical modeling approach in this area of research. Moreover, in our country this clinical biomechanical approach is relatively new and the interaction with physicians and the establishment of common research projects is now starting to happen.

8.4 References

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Chapter 9

9 Publications

9.1 Current work publications and communications

Parts of this dissertation have been published or submitted for publication with first and co-authorship. They have also been presented in congresses and published in abstract books of scientific journals.

Peer-reviewed papers in international journals (ISI) with first authorship

João, Filipa; Veloso, António. 2013. "Quantifying the effect of plyometric hopping exercises on the musculoskeletal system: contributions of the lower limb joint moments of force to ground reaction forces in hopping exercise",*Journal of Mechanics in Medicine and Biology* 13, 1: 1350027 - 1350027. doi: 10.1142/S0219519413500279.

Filipa João, António Veloso, Sílvia Cabral, Vera Moniz-Pereira, Thomas Kepple. "Synergistic interaction between ankle and knee during hopping revealed through induced acceleration analysis." Submitted to the journal *Human Movement Science* (under review).

Filipa João, António Veloso. "Muscle tendon behavior in single-leg cyclic hopping". Submitted to the journal *Computer Methods in Biomechanics and Biomedical Engineering* (under review).

Filipa João, Sérgio Alves, Rita Pereira, José Alves, Mario Secca, Michael Noseworthy, Cristina Menezes, António Veloso. "Assessment of lower leg muscles architecture using diffusion tensor MRI: changes with cyclic stretch-shortening exercise". Submitted to the *Journal of Science and Medicine in Sport* (under review).

Filipa João, António Veloso. "Neuromechanical changes in cyclic hopping exercise". Submitted to the *European Journal of Sport Science* (under review).

Congresses Proceedings

João, Filipa; Veloso, António P; Cabral, Sílvia; Pereira, Vera M; Kepple, Thomas. 2013. "Contribuição dos momentos de força do membro inferior no salto unilateral (hopping)", Trabalho apresentado em 5º Congresso Nacional de Biomecânica, In Atas do 5º Congresso Nacional de Biomecânica, Espinho.

Alves, Sérgio; Pereira, Ana R; Alves, José N; João, Filipa; Secca, Mário F; Veloso, António P. 2013. "Avaliação do efeito do exercício intenso na estrutura e arquitetura de vários músculos da perna através de tensores de difusão em ressonância magnética", Trabalho apresentado em 5º Congresso Nacional de Biomecânica, In Atas do 5º Congresso Nacional de Biomecânica, Espinho.

João, Filipa; Veloso, António P. 2012. "MUSCLE NET MOMENT CONTRIBUTION IN REPETITIVE HOPPING EXERCISE", Trabalho apresentado em 18th Congress of the European Society of Biomechanics, 1-4 July 2012, In 18th Congress of the European Society of Biomechanics, 1-4 July 2012, Lisboa.

Secca, Mário F; Alves, Sérgio; Noseworthy, Michael; João, Filipa; Veloso, António. 2012. "RECOVERY OF LEG MUSCLE ADC AND FA AFTER FATIGUE (WORK IN PROGRESS)", *Journal of Biomechanics* 45, S1: S457 - S457.

João, Filipa; Veloso, António P. 2011. "RELATIVE CONTRIBUTION OF THE JOINT MOMENTS TO THE ANGULAR ACCELERATION OF THE LOWER LIMB IN HOPPING EXERCISE". *Biomechanics in Sports - Portuguese Journal of Sport Sciences* 11, 2: 801 - 804.

Secca, Mário F; Alves, Sérgio; João, Filipa; Noseworthy, Michael D; Veloso, António P. 2012. "Recovery of leg muscle ADC and FA after fatigue (Work in Progress)", Trabalho apresentado em 18th Congress of the European Society of Biomechanics, 1-4 July 2012, In 18th Congress of the European Society of Biomechanics, 1-4 July 2012, Lisboa.

Alves, Sérgio; Pereira, Ana R; Alves, José N; Secca, Mário F; João, Filipa; Veloso, António P; Noseworthy, Michael D; Menezes, Cristina. 2012. "The assessment of the effect of fatigue on muscle architecture through Diffusion Tensor Imaging (DTI) parameters", Trabalho apresentado em 29th Annual Scientific Meeting of the European Society for Magnetic Resonance in Medicine and Biology, In Proceedings of the 29th Annual Scientific Meeting of the ESMRMB, Lisboa.

João, Filipa; Veloso, António P. 2011. "RELATIVE CONTRIBUTION OF THE JOINT MOMENTS TO THE ANGULAR ACCELERATION OF THE LOWER LIMB IN HOPPING EXERCISE", Trabalho apresentado em 29th Annual Conference of the International Society of Biomechanics in Sports, In Biomechanics in Sports - Portuguese Journal of Sport Sciences, Porto.

João, Filipa; Veloso, António P. 2011. "MUSCLE CONTRIBUTION TO BRAKING AND PROPULSION DURING CYCLIC HOPPING EXERCISE", Trabalho apresentado em XXIII Congress of the

International Society of Biomechanics, In Proceedings of the XXIII Congress of the International Society of Biomechanics, Bruxelas.

Alves, Sérgio; João, Filipa; Secca, Mário F; Noseworthy, Michael D; Veloso, António P; Menezes, Cristina. 2011. "Recuperación del ADC y FA del músculo de la pierna después de la fatiga - Resultados preliminares", Trabalho apresentado em V Congreso Latinoamericano de Ingeniería Biomédica, In Actas do V Congreso Latinoamericano de Ingeniería Biomédica, Habana.

Alves, Sérgio; Secca, Mário F; João, Filipa; Veloso, António P; Noseworthy, Michael D; Menezes, Cristina. 2011. "RECUPERAÇÃO DA ANISOTROPIA FRACCIONAL E COEFICIENTE DE DIFUSÃO APARENTE DO SOLEAR E TIBIAL ANTERIOR APÓS O EXERCÍCIO (WORK IN PROGRESS)", Trabalho apresentado em 4º Congresso Nacional de Biomecânica, In Luis Roseiro & Augusta Neto. (Eds.) Actas do 4º Congresso Nacional de Biomecânica, Coimbra.

João, Filipa; Veloso, António P. 2010. "ESTIMATION OF CHANGES IN MUSCLE ACTIVATION ASSOCIATED TO FATIGUE USING STATIC OPTIMIZATION", Trabalho apresentado em International Union of Theoretical and applied Mechanics Symposium - Analysis and Simulation of Human Movement, In Proceedings of the IUTAM Symposium - Analysis and Simulation of Human Movement, Leuven.

João, Filipa; Veloso, António P; Pereira, Vera M; Agostinho, Ruth. 2009. "HOW DOES EXHAUSTIVE CYCLIC LOADING EXERCISE AFFECT MUSCLE ACTIVATION AND LOWER LIMB JOINT MOMENTS OF FORCE?", Trabalho apresentado em 22th Congress of the International Society of Biomechanics, In Proceedings of the 22th Congress of the International Society of Biomechanics, Cape Town.

João, Filipa; Veloso, António P; Pereira, Vera M; Agostinho, Ruth. 2009. "THE INFLUENCE OF EXHAUSTING SSC EXERCISE ON THE LOWER LIMB NEUROMECHANICAL BEHAVIOUR (PRELIMINARY STUDY)", Trabalho apresentado em 14th Annual Congress of the European College of Sports Sciences, In Proceedings of the 14th Annual Congress of the European College of Sports Sciences, Oslo.

João, Filipa; Veloso, António P; Pereira, Vera M; Agostinho, Ruth. 2009. "ALTERAÇÕES DINÂMICAS DO SISTEMA MUSCULO-ESQUELÉTICO ASSOCIADAS À FADIGA. (ESTUDO PRELIMINAR)", Trabalho apresentado em 3º Congresso Nacional de Biomecânica, In Actas do 3º Congresso Nacional de Biomecânica, Bragança.

Communications at scientific meetings

Oral Communications

João, Filipa; Veloso, António P; Cabral, Sílvia; Pereira, Vera M; Kepple, Thomas. Contribuição dos momentos de força do membro inferior no salto unilateral, 5º Congresso Nacional de Biomecânica, Espinho, 2013.

João, Filipa; Veloso António P. MUSCLE NET MOMENT CONTRIBUTION IN REPETITIVE HOPPING EXERCISE, 18th Congress of the European Society of Biomechanics, Lisbon, 2012.

João, Filipa; Veloso, António P. RELATIVE CONTRIBUTION OF THE JOINT MOMENTS TO THE ANGULAR ACCELERATION OF THE LOWER LIMB IN HOPPING EXERCISE,24th Congress of the International Society of Biomechanics, Porto, 2011.

João, Filipa; Veloso, António P; Pereira, Vera M; Agostinho, Ruth. ALTERAÇÕES DINÂMICAS DO SISTEMA MUSCULO-ESQUELÉTICO ASSOCIADAS À FADIGA. (ESTUDO PRELIMINAR), 3º Congresso Nacional de Biomecânica, Bragança, 2009.

Poster Communications

João, Filipa; Veloso, António P. MUSCLE CONTRIBUTION TO BRAKING AND PROPULSION DURING CYCLIC HOPPING EXERCISE, XXIII Congress of the International Society of Biomechanics, Bruxelas, 2011.

João, Filipa; Veloso, António P. ESTIMATION OF CHANGES IN MUSCLE ACTIVATION ASSOCIATED TO FATIGUE USING STATIC OPTIMIZATION,IUTAM Symposium - Analysis and Simulation of Human Movement, Leuven, 2010.

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9.2 Other publications in journals (ISI) as first author or with co-authorship

1. João F, Amado S, et al. (2010). "Anatomical Reference Frame versus Planar Analysis: Implications for the Kinematics of the Rat Hindlimb during Locomotion." Reviews in the Neurosciences **21**(6): 469-485.

Abstract

Functional recovery is the primary goal of therapeutic intervention in neuromuscular rehabilitation. The purpose of this study was to perform a segmental kinematic analysis using both planar angles computation and a tridimensional (3D) reconstruction of the rat hindlimb, regarding the morphology and the movement of each segment. Seven rats were evaluated for natural overground walking, and motion capture of the right hindlimb was collected with an optoeletronic system while the animals walked in the track. 3D biomechanical analyses were carried out and hip, knee, ankle, and metatarsophalangeal joint angular displacements were calculated. For flexion/extension, the knee joint and toe segment were statistically different between planar and 3D analysis, with the toe segment performing less extension at initial contact (IC) and the amplitude during swing phase for the knee being larger. During abduction/adduction, all hip joint parameters were statistically different except at IC and toe-off (TO) instants, the planar angles being higher than the 3D angles. In the horizontal plane, significant differences were found for ankle peaks of rotation, with increased results for the planar angles. In conclusion, a comparison between planar and 3D segmental kinematic analysis using a tridimensional reconstruction of the rat hindlimb demonstrated that different joints have different motion patterns within motion planes, probably related with physiological constraints and muscle actions. A major indication of the need for an anatomical reference frame kinematic analysis is supported by the knowledge that neuromuscular diseases are related to important clinical signs or motor deficits that should be observed, qualified, and quantified.

2. Quaresma C, **João F**, Fonseca M, Secca M F, Veloso AP, O'Neill J G, Branco J. (2010). "Comparative evaluation of the tridimensional spine position measured with a new instrument (Vertebral Metrics) and an optoelectronic system of stereophotogrammetry." Med Biol Eng Comput **48**(11): 1161-1164.

Abstract

We designed and built a non-invasive instrument, called Vertebral Metrics, to measure the x, y, and z positions of each spinous process of the spine on a standing position. In the present study, we perform a comparative evaluation of Vertebral Metrics, by comparing the results obtained from this instrument with those from a validated optoelectronic system of stereophotogrammetry, with 10 infrared cameras. The sample was composed of 11 women aged between 14 and 39 years. After marking the various points on the spinal column, from the first cervical vertebra to the first sacral vertebra, they were measured first with the new instrument (Vertebral Metrics) and then by means of the optoelectronics system. Afterwards, the results were subjected to a thorough comparison. The statistical comparison of the results was performed using an ANOVA model with three factors (Instrument, Subject, and Vertebra) for the intervertebral distance.

3. Amado S, Armada-da-Silva P A, **Joao F**, Mauricio A C, Luis A L, Simoes M J, Veloso A P (2011). "The sensitivity of two-dimensional hindlimb joint kinematics analysis in assessing functional recovery in rats after sciatic nerve crush". Behav Brain Res **225**(2): 562-573.

Abstract

Walking analysis in the rat is increasingly used to assess functional recovery after peripheral nerve injury. Here we assess the sensitivity and specificity of hindlimb joint kinematics measures during the rat gait early after sciatic nerve crush injury (DEN), after twelve weeks of recovery (REINN) and in sham-operated controls (Sham) using discriminant analysis. The analysis addressed gait spatiotemporal variables and hip, knee and ankle angle and angular velocity measures during the entire walking cycle. In DEN animals, changes affected all studied joints plus spatiotemporal parameters of gait. Both the spatiotemporal and ankle kinematics parameters recovered to normality within twelve weeks. At this time point, some hip and knee kinematics values were still abnormal when compared to sham controls. Discriminant models based on hip, knee and ankle kinematics displayed maximal sensitivity to identify DEN animals. However, the discriminant models based on spatiotemporal and ankle kinematics data showed a poor performance when assigning animals to the REINN and Sham groups. Models using hip and knee kinematics during walking showed the best sensitivity to recognize the reinnervated animals. The model construed on the basis of hip joint kinematics was the one combining highest sensitivity with robustness and high specificity. It is concluded that ankle joint kinematics fails in detecting minor functional deficits after long term recovery from sciatic

nerve crush and extending the kinematic analysis during walking to the hip and knee joints improves the sensitivity of this functional test.

4. Wangdo Kim, **Filipa João**, John Tan, Patricia Mota, Veronica Vleck, Liliana Aguiar and Antonio Veloso (2013). "The natural shock absorption of the leg spring". J Biomechanics **46(1)**:129-36.

Abstract

When a human being runs, muscles, tendons, and ligaments together behave like a single linear spring. This "leg spring" can be described remarkably well by spring/mass models. Although leg-stiffness during running (and logically, therefore, in hopping) has been shown to be adjusted in line with the individual characteristics of the external contact surface, the relative contribution of each of the sub-components of the leg spring to the mechanics of running is unclear. The aim of this study is to investigate how the natural (third order) vibration of a lightly damped human leg spring can act as a shock absorber, at the moment of foot-floor impact, in hopping exercise. We found that if the leg is displaced from a position of equilibrium by a natural vibration about one of harmonic modes, then a vibration about this harmonic mode evokes a system of forces in the leg spring which in its turn tends to produce a motion on the original harmonic mode. Our results indicate that if the control of shock absorption of the leg spring, distributed over the subject and environment, was inhibited arbitrarily, it could not vibrate at its own natural mode hence will reduce the natural shock absorption ability.

5. Kim, W., Veloso, A. P., Araujo, D., Vleck, V., & **Joao, F.** (2013). "An informational framework to predict reaction of constraints using a reciprocally connected knee model". Comput Methods Biomech Biomed Engin. **4**: 1-12.

Abstract

Researchers have used screw theory to describe the motion of the knee in terms of instantaneous axes of the knee (IAK). However, how geometric change to the dynamic alignment of IAK may affect stance phase of foot loading has not yet been fully explained. We have tested our informational framework through readily accessible benchmark data (Fregly et al. 2012): muscle contraction and ground reaction force are compounded into a wrench that is reciprocal to the IAK and resolved into component wrenches belonging to the reciprocal screw system. This revealed the special screw system that defines the freedom available to the knee

and more precisely revealed how to measure this first order of freedom. After this step, we determined the reciprocal screw system, which involves the theory of equilibrium. Hence, a screw system of the first order will have a screw system of the fifth order as its reciprocal. We established a framework the estimation of reaction of constraints about the knee using a process that is simplified by the judicious generation of IAK for the first order of freedom in equilibrium.

Chapter 10

10 Appendix