SÃO PAULO STATE UNIVERSITY
“JÚLIO DE MESQUITA FILHO”
INSTITUTE OF BIOSCIENCE - RIO CLARO

GRADUATE PROGRAM IN CIÊNCIAS DA MOTRICIDADE

CO-ORDINATION AND FATIGUE OF COUNTERMOVEMENT JUMP

GLEBER PEREIRA

Thesis submitted in partial fulfillment of the requirements of the Institute of Bioscience of Rio Claro campus, Sao Paulo State University, for the degree of Doctor in Ciências da Motricidade.

Rio Claro
July 2007

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Co-advisor: José Angelo Barela, Ph.D.

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"It is quite difficult to understand fatigue

from single muscle fibre to whole body."

Brenda Bigland-Ritchie

I dedicate this thesis to my family

and Eduardo Kokubun, Ph.D.

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ABSTRACT

This thesis explored the cause of fatigue in countermovement jump performed under different intensities and its influence upon movement pattern. These aims were tested using two experimental studies. The first one aimed to compare the causes of fatigue development during a short- (=10 min) and a long-term (=20 min) countermovement jump protocols through the manipulation of resting interval. The results indicated that after short-term fatigue protocol, maximal voluntary isometric contraction reduction was accompanied by central and peripheral fatigue. On the other hand, after long-term fatigue protocol, maximal voluntary isometric contraction reduction was accompanied by peripheral fatigue. The second study used similar exercise protocol to induce fatigue on neuromuscular system. It was aimed to investigate the movement pattern of countermovement jump throughout fatigue exercise protocols. The timing and sequencing of inter-segmental movement and of muscle activation remained relatively unaltered under fatigue. However, in order to sustain required jump height, few adjustments were performed on control and on coordination of fatigued jumps, such as increased joint flexion and earlier initiation of joint extension. This result was quite similar irrespective of fatigue-induced exercise intensities. Considering both experimental studies, the results of present thesis suggest that the cause of fatigue in countermovement jump is task-dependent in which depends on what fatigue protocol is used, thereby the cause of fatigue can move from one site to another. Even with the neuromuscular system impaired, there was not change on the movement pattern of countermovement jump, irrespective of exercise duration and cause of fatigue. However, adjustments on control and coordination of countermovement jump were performed in order to compensate the force loss and to maintain the targeted jump height.

Keywords: central fatigue, peripheral fatigue, coordination of movement, control of movement

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CHAPTER 1

General introduction

1.1 Introduction

When a person is performing any motor task in sport activities we always think that he/she is able to accomplish the required goal with no physical limitations. However, intensity and duration of exercises can inhibit differently the integrity of the central nervous system and its sub-systems, in such a way that limits the performance. This phenomenon is named fatigue and the understanding of its possible causes may help to elaborate exercise training and performance in order to postpone the reversible process of impairment and extend the task. According to different muscle type contraction, exercise intensity and mode of exercises, there are two causes of fatigue. The causes of fatigue can be central and peripheral. For instance, stretch-shortening cycle muscle contraction involved in running and vertical jumping (e.g., countermovement jump) respond differently to exercise duration which can lead to central and/or peripheral fatigue. A direct comparison between short and long exercise duration can enhance the knowledge on central and on peripheral causes of fatigue in stretch-shortening cycle exercise.

Although countermovement jumps are frequently used in volleyball and basketball sports and it seems to involve low motor complexity, the success depends on how the many joints and muscle activations are organised by the motor system to accomplish the goal. Despite the apparent simplicity, countermovement jumping is a complex motor task in which variations in timing and sequencing of movement and of muscle activation may lead to a reorganisation of coordination. For instance, impairment of the neuromuscular system may cause such reorganisation of coordination in order to compensate the muscle fatigue (van Ingen Schenau *et al.* 1995). Some studies have shown that fatigue plays influence upon movement pattern (Cote *et al.* 2002, Huffenus *et al.* 2006); however, no studies have demonstrated how it reacts to different causes of fatigue.

This thesis explores the causes and consequences of fatigue in countermovement jumps performed at different work to rest ratio.

1.2 Theoretical background

1.2.1 Muscle fatigue

Theoretically, motor tasks executed during sports or daily activities should be maintained infinitely, in order to achieve high level of performance or productivity. However, empirically, muscles cannot sustain activity and, consequently, generate force infinitely. When this inability to generate force occurs, it is formally termed muscle fatigue. However, a number of definitions have been used for fatigue. Fatigue can be defined as a reduction in maximal voluntary effort (Bigland-Ritchie & Woods, 1984). On the other hand, fatigue can be defined as a failure to maintain the required or expected force (Edwards, 1981). Both definitions can be used, if it is understood as the process and as the point of fatigue. An example is an individual performing a submaximal effort. In this case, the maximal force-generating capacity of muscles (assessed from brief maximal voluntary contraction) starts to decline once exercise commences until being equal to the submaximal target effort, i.e., the process of fatigue. Such a process begins almost at the onset of the exercise and develops progressively until the muscles fail to perform the required submaximal task. At the instant that maximal force-generating capacity of muscles is less than the submaximal target effort, the point of fatigue is established because there was a failure to maintain the required or expected force. Therefore, in this thesis, muscle fatigue is interpreted as a physiological process which leads to a reduction in maximal voluntary effort until the point that the required or expected effort is no achievable.

The causes of muscle fatigue have been investigated by many scientists. In a general view, central and peripheral factors have been suggested to be responsible for muscle fatigue. Central fatigue is defined as a progressive reduction in voluntary activation of muscle during exercise (Gandevia, 2001). Voluntary activation can be understood as the level of voluntary drive from the central nervous system to the muscles during an effort. Maximal voluntary activation can be assessed using twitch interpolation (i.e., stimulus is delivered to the motor axons innervating the muscle) during a maximal voluntary contraction (Gandevia, 2001). Peripheral fatigue is produced by changes at or distal to the neuromuscular junction (Gandevia, 2001). Compound muscle action potential and muscle twitch response to a single stimulus of a nerve are useful tools to investigate peripheral fatigue.

In 1891, Angelo Mosso had already distinguished between central and peripheral fatigue suggesting that brain fatigue could reduce the strength of the muscles (Di *et al.*, 2006). This association between central and peripheral fatigue proposed by Mosso still remains intriguing to a large number of researchers. To establish that muscle fatigue has always been caused by a unique process or combination of processes requires that the time course of the process match those of force or power output, both during the period of exercise and also throughout recovery. A second criterion is to establish that this match between force and the process being considered exist under all kinds of exercise conditions (Bigland-Ritchie *et al.*, 1995). Several studies have been performed addressing this question and their results are conflicting which seems to be related to differences in the particular exercise protocols used to generate fatigue. Thus, it was suggested that the amount of stress placed on each site (e.g., calcium release or alpha motoneurons inhibition) depends on the type of exercise, or task, from which fatigue develops. This is termed task-dependency (Bigland-Ritchie *et al.*, 1995). For instance, when a task is changed, any increased stress on one site may be compensated by a relief at other, so that the overall outcome appears to operate like a single process (Enoka & Stuart, 1992). The term "task", used to compare task-related changes with fatigue, including not only the performance of different types of exercise by any given muscle group actions (isometric, dynamic, etc.), but also similar types of exercise performed by various muscles that have different contractile properties and functions.

Regarding to differences in exercise-type, isolated forms of muscle action, such as isometric, concentric and eccentric contractions are not performed during natural human locomotion. Basically, thereafter the muscle is stretched (eccentric), it is shortened (concentric) and this type of muscle action occurs during running, hopping and jumping, for example. This combination of eccentric and concentric actions forms a natural type of muscle function called the stretch-shortening cycle (SSC) (Komi, 2000). In SSC-type muscle actions, the nature of stretches involved in the active braking phase of SSC are usually very fast and of short duration. Neural (i.e., reflex and central neural pathways) and peripheral (mechanical behaviour and structural modifications of the muscle-tendon unit) components play a key role in the optimal muscle activation. In addition, the metabolic loading of SSC exercise may differ considerably from those of pure concentric or eccentric exercises (Nicol *et al.*, 2006). Due to SSC-type muscle function is quite different from isolated forms of muscle function; it becomes an appealing subject to study muscle fatigue.

SSC exercise often results in reversible muscle damage; this is associated with delayed-onset muscle soreness, and with proprioceptive and neuromuscular impairments that may last for several days. After SSC exercises, these neuromuscular impairments are typically associated with acute and delayed changes in muscle mechanics and activation that result in major consequences on joint and muscle stiffness regulation (Komi, 2000). These fatigue-induced performance deteriorations and subsequent long-term recovery usually take place in a bimodal fashion (Nicol *et al.*, 2006). In this bimodality, the acute metabolically induced reduction is followed after a few hours by a short-term recovery, which is in turn followed by a secondary reduction with longer lasting recovery. However, the exact causes (central and peripheral) of neuromechanical parameter changes after SSC fatigue are often difficult to assess because of differences in the protocols used.

Considering fatigue protocols which used either rebound (Kuitunen *et al.*, 2004) or drop jump (Skurvydas *et al.*, 2000), and performed during either short- [e.g., = 10 min and/or ~ 100 jumps] (Avela *et al.*, 2006) or long-term [e.g., ~ 25 min and/or = 200 jumps]

(Regueme *et al.*, 2005) exercise, three main aspects can be highlighted to differentiate them: (i) jumping height, (ii) interrupting test, and (iii) resting interval. In the first aspect, related to jump target height, participants have been asked to jump either maximally (Strojnik & Komi, 1998), or submaximally [from 60 to 80%] (Nicol *et al.*, 1996). The second aspect is related to test interruption: after predetermining work is accomplished [from 50 to 100 jumps] (Hortobagyi *et al.*, 1991), or after inability to reach the target height (Strojnik & Komi, 2000), or after inability to jump off (Kuitunen *et al.* 2004). The third aspect relates to resting interval, more specifically the interval between jumps (Skurvydas *et al.*, 2002), or between jumping sets (Avela *et al.*, 2006), or with no resting interval [i.e., performed consecutively] (Kuitunen *et al.*, 2004).

In the fatigue protocols aforementioned, interestingly, central and peripheral factors may cause deficit on maximal voluntary contraction even when a parameter of such protocols has changed. For instance, Regueme *et al.* (2005) performed submaximal rebound tests and the reduced maximal isometric voluntary contraction was associated with muscle contractile impairment. On the other hand, Kuitunen *et al.* (2002) performed maximal rebound tests and the greater maximal isometric voluntary contraction deficit was associated with the larger voluntary activation deficit. Therefore, the hypothesis of task-dependency to explain differences on fatigue causes seems to be suitable.

To explore the cause of fatigue in SSC-type muscle action, an identical fatigue protocol, except changing one parameter, seems to be interesting in order to investigate fatigue in intermittent countermovement jumps performed during short- and long-term exercise, getting to a direct comparison. In SSC-fatigued exercise protocols, intermittent countermovement jumping is useful due to the ease work which performance characteristics can be manipulated or controlled and due to its ecological validity to a large number of sports tasks. Therefore, using intermittent countermovement jumps performed under different work to rest ratio, manipulating resting interval between jumps as performed by Pereira (Pereira *et al.*, 2004), makes possible a direct comparison between short and long exercise duration to enhance the knowledge on central and peripheral causes of fatigue in SSC exercise.

1.2.2 Control and coordination of movement

Bernstein (1967) stressed the richness of human movement in terms of the plethora of directions and types of mobility (degrees of freedom) that can be performed. Despite the abundance of movement possibilities, humans lead to perform motor tasks in a coordinated and controlled form using relatively few degrees of freedom. In addition, the motor system has a number of characteristics that make the control of complex movement far from simple (van Ingen Schenau *et al.*, 1995). Such characteristics are tendon elasticity, muscle fiber characteristics and interaction between body segments. Having the complexity of the locomotor apparatus and the number of possibilities to execute a single movement in mind, Bernstein questioned how the central nervous system could be responsible for the control of all motor actions through inputs sent to each motor unit. This questioning has been termed as the degrees of freedom problem.

The second problem identified by Bernstein (1967) related to the context conditioned variability in which movements are performed. In this case, the environment influences the organization and the execution of movements and the environment conditions change across movement performances. For instance, similar muscle neural drive can produce different movements and different muscle neural drive can produce similar movements (Turvey, 1990). Once more, Bernstein questioned how central nervous system could be responsible to control all components of the motor actions.

Concomitantly to Bernstein's idea related to context conditioned variability, it was emphasised the needed for a mutual relationship between animal and environment (Gibson, 1979). According to Gibson (1979), the animal suffers influence from the environment and they are complementary in order to obtain (the animal) and to provide (the environment) all the characteristics that are necessary for perception of the context. Considering Bernstein and Gibson's ideas, they questioned many assumptions about movement control and pushed many researchers to search for different assumptions and explanations about control and coordination of movement. Such explanation is based on a model of the motor system that should be robust to organize the many degrees of freedom. At the same time, the motor system should be sensitive and flexible to changes due to the context which the movement is inserted. From this view of the motor system, a dynamical approach to the control and coordination of the movement execution has been suggested (see further in Kelso *et al.*, 1980, Kugler *et al.*, 1980) and many researchers are still working on many issues regarding motor control under a dynamical view. Basically, the solution to the degrees of freedom problem was that muscles are controlled in a synergy form, instead of individual form. Thus, the system would constrain the functioning of many muscles leading them to act as a functional unit, named a coordinative structure.

Bernstein (1967) purposed that coordination is overcoming excessive degrees of freedom of our movement organs, that is, turning the movement organs into a controllable system. Kugler *et al.* (1980) had built on Bernstein's formulation and suggested a distinction between the term coordination and control. Briefly, coordination is the function that constrains the potentially free variables into a behavioural unit. The basis of this function is a set of variables:

$$(A, B, C, \ldots X, Y, Z)$$

These variables may be constrained into a coordination function:

$$f\,(A, B, C, \ldots X, Y, Z)$$

Control is the process by which values are assigned to the variables in the function, i.e., parameterizing the function:

$$f\,(A_i, B_j, C_k, \ldots X_r, Y_s, Z_t)$$

In general, the coordination pattern has been approached through timing and sequencing of segmental movement (Clark *et al.*, 1989, Hudson, 1986). Obviously, the muscle activation is responsible to the body segment movement and it is measured through electromyography. Hence, the timing of muscle activation is considered coordination, whilst its magnitude is considered control (Lay *et al.*, 2002). In short, relationship between two or more parts (or variables) is considered coordination, and magnitude of signal for any variable is considered control.

Newell (1986) hypothesized that controlled and coordinated movements are dependent on the relationship among the performer (organism), the environment and the desirable task. These three groups of influences on movement execution would be understood as constraints. Specifically, he defined constraint as boundaries or features that limit the motion under consideration, which in engineering parlance means that constraints reduce the number of possible configurations of a system. Then, Newell (1986) proposed that there are three categories of constraints that interact to determine for a given organism the optimal pattern of coordination and control for any activity. These are organismic constraints, environmental constraints and task constraints (see figure 1), which are explained below according to Newell’s view.

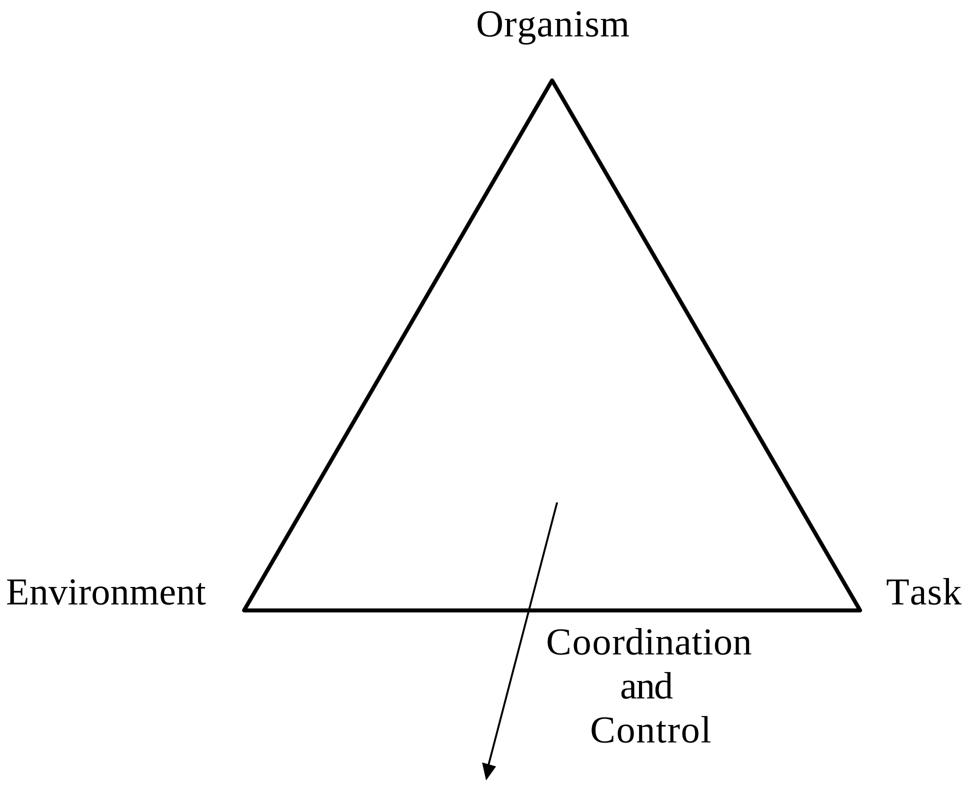


Figure 1.1: A schematic diagram of the categories of constraints that specify the optimal pattern of coordination and control (Newell, 1986).

The organismic constraint resides at each level of analysis of the organism, typically interpreted as structural and functional constraints. Body weight, height and shape are assumed to be structural constraints. In contrast, joint amplitudes are often taken as a functional constraint. Environmental constraints are generally recognised as those ones that are external to the organism and are usually not manipulated by the experimenter (e.g., gravity). In general, the focus of task constraints is the goal of the activity and the specific constraints imposed. The motor system searches for a stable pattern of coordination and control that accommodates the prevailing constraints (Newell, 1986). However, most tasks do not specify the pattern of coordination utilised by the performer, consequently, a new pattern of coordination to satisfy the outcome of the act may emerge.

Based on the dynamical approach, schematized on figure 1, scientists have conducted many studies to understand how the human organism responds and interacts to different tasks and environments constraints. From daily activities to sports performance, the maintenance of the ability to achieve the goal of the task, in any environment, is always aimed. However, almost at the onset of exercise the process of fatigue starts to influence the neuromuscular system, and consequently as the exercise is maintained, it is hard to accomplish task goal. On the one hand, a new coordination pattern may emerge in order to compensate the force loss and to sustain the performance (van Ingen Schenau *et al.*, 1995). On the other hand, the coordination pattern may be deteriorated even though the force loss has enhanced. From this perspective, fatigue can be understood as a time dependent organismic constraint. Therefore, the longer the duration of the motor task the greater is the level of fatigue throughout constant exercise intensity. Despite this logical thought, very little is known about the effects of fatigue on the coordination pattern of motor activities.

Studies have investigated the effects of muscle fatigue upon control and coordination pattern of multi-joint movement, such as lifting (Sparto *et al.*, 1997), throwing (Forestier & Nougier, 1998, Huffenus *et al.*, 2006), cycling (Hautier *et al.*, 2000), sawing (Cote *et al.*, 2002), and vertical jumping (Rodacki *et al.*, 2001, Rodacki *et al.*, 2002). In general, these studies have reported that the motor system has changed inter-segmental organization and timing of movement, muscle activation timing and magnitude, and joint contribution in order to compensate for the effects fatigue. Hence, fatigue interpreted as a constraint is an appealing way to understand the coordination and control of movement performed under this reversible physiological state.

Vertical jumping is a useful motor task to investigate the fatigue effects on movement pattern. Firstly, vertical jump is a multi-joint task which involves many muscle groups to achieve its goal (jump height). Secondly, in such motor task, the performance excellence depends on to what extent the movement pattern is well performed (Bobbert & Van Soest, 1994). Thirdly, as mentioned previously, vertical jumping is an important motor task involved in many sport activities, such as volleyball and basketball, and it enhances the ecological validity of the studies. Furthermore, another aspect must be pointed. It is related to when movement pattern performed under fatigue is measured in order to compare with non-fatigued movement pattern. This is an important issue due to adjustments in movement pattern which may occur during the process of fatigue (van

Ingen Schenau *et al.*, 1995). Therefore, performing measurements of movement pattern throughout the exercise until the point of fatigue are interesting to investigate.

1.3 Goals and outline of the thesis

The present thesis explores the causes and consequences of fatigue in countermovement jumps performed at different work to rest ratio. This study focused on the following research hypotheses:

1. The cause of fatigue in countermovement jump is different when it is performed at different work to rest ratio;
2. The movement pattern of countermovement jump is (re)organised when it is performed under fatigue.

To explore such hypotheses, using similar exercise protocols, two experimental studies were developed in which countermovement jumps were performed until the point of fatigue. In these studies, the work to rest ratio was manipulated using different rest periods between jumps in order to induce different exercise duration, thereby inducing different number of jumps. The fatigue exercise protocols should last shorter (= 10 min) and longer (= 20 min). Chapters 2 and 3 present both experimental studies.

Chapter 2 compares the causes of fatigue development in countermovement jumps performed during a short- and a long-term protocol. Neuromuscular parameters, such as neuromuscular propagation, mechanical properties of the muscle and maximal voluntary activation capacity, were measured. The outcomes are discussed in terms of central and peripheral influences on muscle fatigue, regarding different work to rest ratio.

Chapter 3 assesses the possible reorganisation of movement pattern of countermovement jumps performed throughout short- and long-term fatigue protocols. Kinematic, kinetic and electromyographic measurements were obtained from all countermovement jumps. The results are related to the effects of fatigue-induced protocols on movement pattern.

In the **chapter 4** the findings with respect to the cause of fatigue on countermovement jump are discussed in association with the outcomes of movement pattern. General conclusions and recommendations for future studies are provided, as well.

CHAPTER 2

The causes of fatigue in countermovement jumps

This chapter is under review with the title: *“Neuromuscular parameters after stretch-shortening cycle fatiguing protocols with different rest periods”, in Journal of Applied Physiology.*

2.1. Introduction

Active muscles undergo a combination of stretching (eccentric) and shortening (concentric) contractions during most locomotor activities (e.g., running and jumping). This combination forms a natural type of muscle functioning called the stretch-shortening cycle (SSC) (Komi, 2000). Repetitive SSC (e.g., jumping) are performed in training and competition for many sports (e.g., volleyball, basketball, football) and, depending on the work to rest ratio and number of SSC-type actions, may lead to fatigue. Thus, depending on these parameters, central and peripheral mechanisms play a key role in fatigue development and performance maintenance (Nicol *et al.*, 2003). Understanding the underlying mechanisms of fatigue development for movements involving maximal or near maximal SSC actions with different work to rest ratio may be critical to use better training strategies for such sports.

After short-term SSC exercises (e.g., = 10 min and/or ~100 jumps), decreased force production has been shown to be caused by peripheral mechanisms such as impairment of muscle contractile capacity (Skurvydas *et al.*, 2000, Strojnik & Komi, 1998). On the other hand, force reduction after long-term SSC activities (i.e., ~25 min and/or =200 jumps) has been suggested to be more related to central factors such as reduced levels of voluntary drive (Kuitunen *et al.*, 2004, Nicol *et al.*, 2003). A direct comparison of the underlying mechanisms is not easy due to a large variety of fatiguing protocols applied between short- and long-term exercise protocols. For instance, several authors used short-term maximal drop jump protocols (Avela *et al.*, 2006, Hortobagyi *et al.*, 1991, Skurvydas *et al.*, 2002), while others applied maximal (Strojnik & Komi, 1998) and sub-maximal rebound jumps (Horita *et al.*, 1996, Strojnik & Komi, 2000). Kuitunen *et* *al.* (2002) investigated long-term effects by imposing exhaustive protocols (i.e., 100 maximal drop jumps followed by submaximal rebound jumps), while Regueme *et al.* (2005) analyzed only sub-maximal rebound jumps.

Manipulating the work to rest ratio of maximal jumps is an appealing way to compare the mechanisms of fatigue during short- and long-term exercise protocols using the same task. Smaller work to rest ratio will promote an earlier fatigue in repetitive jumping performance maintenance, whilst greater work to rest ratio may enable participants to perform greater number of jumping. We hypothesized that work to rest ratio may increase the metabolic demand, for the same work, due to a shorter rest period between actions to recover energy supply, having fatigue caused by peripheral mechanisms. On the other hand, greater work to rest ratio may reduce the metabolic demand, for the same work, due to a greater rest period between actions to recover energy supply, having voluntary activation deficit because of greater number of SSC actions. Studies have not tried a direct comparison between jumping protocols, distinct work to rest ratio, producing distinct number of SSC actions (exercise durations). Therefore, the purpose of this study is to compare the causes of fatigue during a short- and a long-term countermovement jump protocols through the manipulation of work to rest ratio.

2.2. Methods

Participants. Eleven healthy males volunteers (mean ± SD; 20.7 ± 2.8 years; 77.1 ± 8.5 kg; 1.85 ± 9 m) participated in this study. They practiced jumping and running activities recreationally at least twice a week. Participants were asked to refrain from caffeine ingestion and severe physical activity in the day before the experimental session. This study was approved by the research ethics committee of Manchester Metropolitan University and was conducted according to the Declaration of Helsinki. Participants were informed about the procedures and risks before giving written consent.

Experimental Design. Participants performed four experimental sessions across 21 days. Sessions were at least four days apart to avoid cumulative effect of fatigue. The first experimental session familiarized individuals to the study procedures. Initially in this session, maximal jump height was determined. It was followed by neuromuscular measurements (pre-), fatigue protocol and neuromuscular measurements (post-). Such sequence was maintained during the second and third sessions. The neuromuscular measurements are schematized in Figure 2.1. The fatigue protocol consisted to participants jumping intermittently until the point of fatigue having 6 s jump between jumps. In the second and third experimental sessions the short and long fatigue protocols were performed in a randomly assigned order. The fatigue protocols were set up to allow a short (= 10 min and/or ~100 jumps) and a long test (~25 min and/or ~200 jumps) until fatigue, by fixing jumping height and manipulating the work to rest ratio through different resting interval between jumps. In order to allow similar fatigue level among individuals, each participant jumped with their own specific resting interval. Such resting intervals were adopted according to the number of SSC actions performed in the familiarization section. For instance, a participant who performed 90 jumps having 6 s of resting interval in the familiarization session had his resting interval maintained to short session and increased to long session. These procedures were based on previous study (Pereira 2004). In the fourth experimental session (control session), pre- and post-measurements were separated by a 10 min rest period, in order to investigate whether pre- influence on post-measurements. During all procedures, the participants received consistent verbal encouragement.

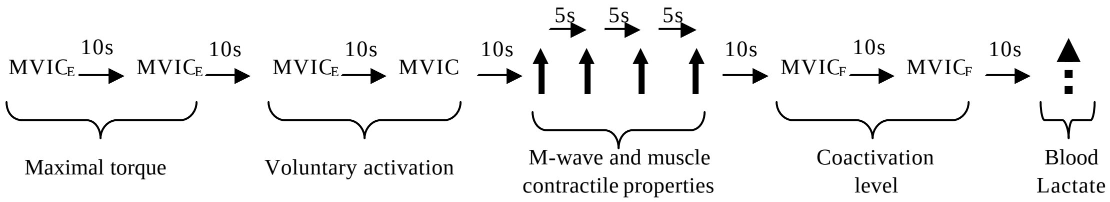


Figure 2.1: Overview of the pre- and post-measurements performed throughout the sessions. The maximal knee extension torque was determined through maximal voluntary isometric contraction (MVIC$_E$); two more MVIC$_E$ were used to estimate the voluntary activation level of quadriceps muscle; four single stimuli (vertical arrows) were applied on femoral nerve to determine M-wave response of vastus lateralis muscle and quadriceps muscle contractile properties; two maximal voluntary isometric knee flexion torque (MVIC$_F$) were used to quantify the coactivation level of biceps femoris muscle during the MVIC$_E$; and blood sample from earlobe to determine blood lactate concentration.

Countermovement Jump. During the familiarization session, participants performed five maximal countermovement jumps (CMJs) with their arms crossed on their chest. Maximal jumps were performed on a force platform (Kistler 9286A, Wintherthur, Switzerland) sampling at a frequency of 1000 Hz. Vertical ground reaction force (vGRF) was smoothed using a zero lag fourth order low pass digital Butterworth filter. Cut-off frequency was defined by residual analysis and set at 13 Hz (Winter, 2005). The net impulse was calculated from the vGRF – time plot through the Matlab program (Math Works, Inc., Natick, Massachusetts, USA), that used the trapezoidal integration approach (Appendix A). The initiation of movement was determined as the instant vGRF decreased by more than 5% of body weight; whilst takeoff instant was determined by the instant at which vGRF became lower than 1% of body weight. Dividing net impulse by body mass, it was determined takeoff velocity (v_{to}) which was used to calculate maximal jump height (h) as follows: $h = v_{to}^2 / (2*g)$. Then, the average of the five maximal vertical jumps was selected as the maximum value. As force platform was used only during maximal jumping, during fatigue protocol, participants were asked to make their head contact with a light weight bar set at 95% jump height plus the participant's tiptoe-standing height, assuming that participants left the ground in tiptoe-standing height. The instant of fatigue was characterized as three consecutive failures to reach the required height. Such instant (total time throughout the session) and the total number of CMJs performed were recorded.

Maximal Voluntary Isometric Contraction (MVIC). An isokinetic dynamometer (Cybex Norm, New York, USA) was used to measure both isometric knee extension ($MVIC_E$) and knee flexion torque ($MVIC_F$). Participants were seated in the chair of the dynamometer with the hip angle at 90º. Knee angle was fixed at 90° of flexion (full knee extension being 0°). Participants' chest and hips were secured by the dynamometer straps to avoid extraneous movements. The estimated center of rotation of the knee joint was aligned with the dynamometer center of rotation at an angle of 90° from the horizontal. Participants' settings on the dynamometer were recorded to guarantee the same position between testing sessions. They were instructed to increase torque so that the maximum was reached after approximately 1-2s with maximum exertions maintained for ~ 3-4 s. The MVICs were separated by 30s rest. All measurements were performed using the right leg. Two MVICs were performed for both knee extensors and knee flexors, respectively. The highest torque value was considered for further analysis.

Electromyographic activity. EMG activity was recorded from the Vastus Lateralis (VL) and the Biceps Femoris (BF) muscles, through two pairs of Ag-AgCl bipolar surface electrodes (Neuroline, Medicotest Denmark). Two electrodes were placed along the sagittal plane over the muscle belly, with an interelectrode distance of 10-mm. Reference electrodes were placed over the right lateral epicondyle of the femur. Before placement of the electrodes, the skin was shaved, lightly rubbed with abrasive gel and cleaned with alcohol swabs to reduce interelectrode impedance to below 5 KO. The EMG signals were differentially amplified with a gain of 2000 and a bandwidth of 10-500 Hz at –3 dB. Raw EMG activity was recorded with a sampling frequency of 2000 Hz and processed with a multichannel analog-to-digital converter (Acknowledge, Biopac Systems, Santa Barbara, California, USA). Both torque and EMG signals were integrated and displayed simultaneously in real time. A 500 ms window was used to calculate EMG root mean square (RMS) values, around the time of peak knee extension

(RMS_{VL}) and knee flexion (RMS_{BF}) torque. The RMS_{BF} was assessed during a maximum flexion contraction and a linear relationship was assumed between RMS_{BF} and knee flexion torque (Reeves *et al.*, 2004). The level of BF antagonist coactivation was estimated by relating the level of RMS_{BF} when acting as an antagonist during knee extension to the level of EMG activity when the same muscle was acting as an agonist.

Voluntary activation. Two supramaximal doublets (two electrical stimuli with 200 µs pulse width, 10 ms inter-stimulus interval and a maximum of 1000 mA output) were applied percutaneously using an electrical stimulator (model DS7, Digitimer, Welwyn Garden City, UK) with stimulation pads (76 mm x 127 mm, Versastim, Conmed Corp., New York, USA) placed over the distal and proximal parts of the quadriceps to evaluate the ability of participants to activate the knee extensors during isometric contractions. The pads were placed in a position that gave the highest twitch torque to a standard stimulus. At the beginning of each experimental session, stimulus intensity was determined by administering twitches of progressively increasing current. Stimuli were delivered with the participants in a relaxed state, starting from 100 mA, increasing by 50-100 mA until no further increment in twitch force was observed. Then, an additional 50 mA was used to guarantee supra-maximal stimulation. Since we investigated only one joint position, it was used the interpolated twitch technique (ITT) method for determining the voluntary activation (VA) level and in contrast to the Central Activation Ratio method, as activation level assessed using this technique is unaffected by the number of stimuli applied (Bampouras *et al.*, 2006). The VA level was calculated as follows (Merton, 1954): VA (%) = [1 – (superimposed doublet torque / rest doublet torque)] x 100. Two doublets were applied to assess muscle activation level. The first doublet was delivered manually during the plateau phase of an $MVIC_E$ attempt (superimposed doublet). A further doublet was evoked automatically 1.5s after the first, corresponding to complete muscle relaxation (rest doublet). Two MVICs were performed and the contraction with the highest torque was selected for analysis.

Muscle Contractile properties. Percutaneous femoral nerve stimulations were used to elicit compound muscle action potential (M-wave). Single square-wave pulses (200 µs pulse width) were delivered to the femoral nerve through a cathode probe (0.5 cm diameter) pressed in the femoral triangle by the same investigator. Then, an anode pad (76 mm x 127 mm) was located in the gluteal fold opposite the cathode. Stimulation and EMG electrode placements were marked on the participants’ skin with a permanent marker to ensure the same position across sessions. During the pre-fatigue protocol measurements, the stimulus intensity was gradually increased from sub-threshold level up to the point at which an increase in current did not result in further increases in peak twitch torque and in M-wave peak-to-peak amplitude of knee extensor torque. This intensity was further increased by 30% to ensure that it was supra-maximal (supra-maximal current amplitudes ranged from 35 to 82 mA). Then, after 40s, four stimuli interspaced by 5s were delivered to the relaxed muscle. Average values from the four stimulations were used for analysis. The twitch contractile measurements from femoral nerve stimulations were: 1) peak twitch torque (PT); 2) contraction time (CT), i.e., the time from the initial rise in the mechanical response to PT; 3) average rate of twitch torque development (RTD = PT/CT); 4) half relaxation time (½ RT), i.e., the time from PT to half its value; 5) average rate of half twitch torque relaxation (½ RTR). Also, for each twitch response, peak-to-peak amplitude (PPA) and peak-to-peak duration (PPD) of electrically evoked compound action potentials on VL muscle were measured.

Blood lactate. Twenty-five micro-litre blood sample was collected from the earlobe to determine lactate concentration. Blood lactate was analyzed using an electro-chemical device (YSI 1500, Yellow Spring Instruments, Ohio, USA).

Statistics. After normality assurance, differences in neuromuscular variables and blood lactate (the latter variable was not measured in the control condition) between test conditions were tested with mixed models. Condition (short, long and control) and instant (pre- and post-) were considered as fixed factors and participants as a random factor. Resting intervals, number of jumps, and fatigue protocol durations were analyzed with a mixed model having condition as a fixed factor and participant as a random factor. A post-hoc test, with an adjustment by Tukey, was used for multi-comparison purposes. Significance was set at $p < 0.05$. Data are presented as mean ± SD.

2.3. Results

The different resting intervals used during the short and long sessions (short = 6.1 ± 1.6 s and long = 8.6 ± 1.6 s; p < 0.01) promoted different number of countermovement jumps (short = 116 ± 38 and long = 201 ± 69 jumps; p < 0.01) and exercise duration (short = 652.6 ± 141.9 s and long = 1454.3 ± 284.2 s; p < 0.01). Blood lactate concentration increased in the short session (pre- = 0.72 ± 0.34 mmol, post- = 2.1 ± 0.79 mmol; p < 0.01) but not in the long session (pre- = 0.59 ± 0.34 mmol; post- = 1.04 ±

0.49 mmol; $p > 0.05$). In addition, blood lactate concentration post- fatigue protocol was greater in the short than in the long session ($p < 0.01$).

There was no significant interaction between Instant and Condition for MVIC_{E}, however, there was a significant main effect of Instant (Table 2.1), i.e., difference from pre- to post-instant (short = -16%, long = -5%, control = 0%). Further, MVIC_{F} (short = -1 %, long = -1 %, control = -2 %) and coactivation (short = -2 %, long = 0 %, control = -1 %) did not change through sessions ($p > 0.05$).

Voluntary activation decreased (-14 %; $p < 0.01$) at the end of short session (Figure 2.2). There was a significant Instant main effect for RMS_{VL} during the MVIC_{E} (short = -22%, long = -14%, control = -2%) and for CT during twitch torque (short = -22%, long = -13%, control = -1%) (Table 2.1). The PT decreased significantly in both short (-26 %) and long (-32 %) sessions (Table 2.1). Nevertheless, the RTD ($p > 0.05$), ½ RT ($p > 0.05$), ½ RTR ($p > 0.05$), PPA ($p > 0.05$), and PPD ($p > 0.05$) showed no significant difference through the sessions (Table 2.1).

Table 2.1. Strength and neuromuscular characteristics of the knee extensor muscle pre- and post-countermovement jump fatigue with different rest intervals.

	Short		Long		Control	
	Pre-	Post-	Pre-	Post-	Pre-	Post-
MVIC$_E$ (N.m)*	329.3±45.0	276.1 ± 66.3	337.7 ± 77.0	321.4 ± 85.3	322.0 ± 88.5	323.1 ± 82.2
RMS$_{VL}$ (mV)*	0.49 ± 0.27	0.38 ± 0.16	0.57 ± 0.3	0.49 ± 0.18	0.46 ± 0.18	0.45 ± 0.14
PT (N.m)	63.2 ± 1.5^a	46.9 ± 12.6	61.7±10.8^a	42.0 ± 9.5	62.9 ± 15.1	64.8 ± 10.9^b
CT (ms)*	83.8 ± 25.3	64.9 ± 22.1	84.1 ± 22.1	73.4 ± 19.3	94.6 ± 18.3	93.5 ± 18.5
RTD (N.m.ms^{-1})	0.81 ± 0.31	0.73 ± 0.21	0.77 ± 0.23	0.61 ± 0.21	0.70 ± 0.26	0.72 ± 0.21
½ RT (ms)	86.1 ± 20.8	75.4 ± 31.7	83.3 ± 25.9	64.9 ± 23.9	66.7 ± 18.8	74.2 ± 24.7
½ RTR (N.m.ms^{-1})	0.38 ± 0.06	0.39 ± 0.25	0.41 ± 0.17	0.38 ± 0.20	0.50 ± 0.18	0.48 ± 0.17
PPA (mV)	10.9 ± 3.1	9.2 ± 2.3	10.3 ± 3.9	10.1 ± 3.3	9.4 ± 2.4	8.8 ± 2.7
PPD (ms)	7.1 ± 0.6	7.9 ± 3.2	9.3 ± 3.7	8.4 ± 3.2	9.7 ± 3.9	9.4 ± 4.1

Values are means ± SD. Maximal voluntary isometric contraction of knee extension (MVIC$_E$); Root mean square of vastus lateralis (RMS$_{VL}$); Peak twitch torque (PT); Contraction time (CT); Average rate of twitch torque development (RTD); Half relaxation time (½ RT); Half average rate of twitch torque relaxation (½ RTR); Peak-to-peak amplitude (PPA) and peak-to-peak duration (PPD) from M-wave response. *Instant main effect ($p < 0.05$); ^{a}Significantly different from post-instant within and between short and long sessions ($p < 0.05$); ^{b}Significantly different from post-instant in short and in long sessions ($p < 0.01$).

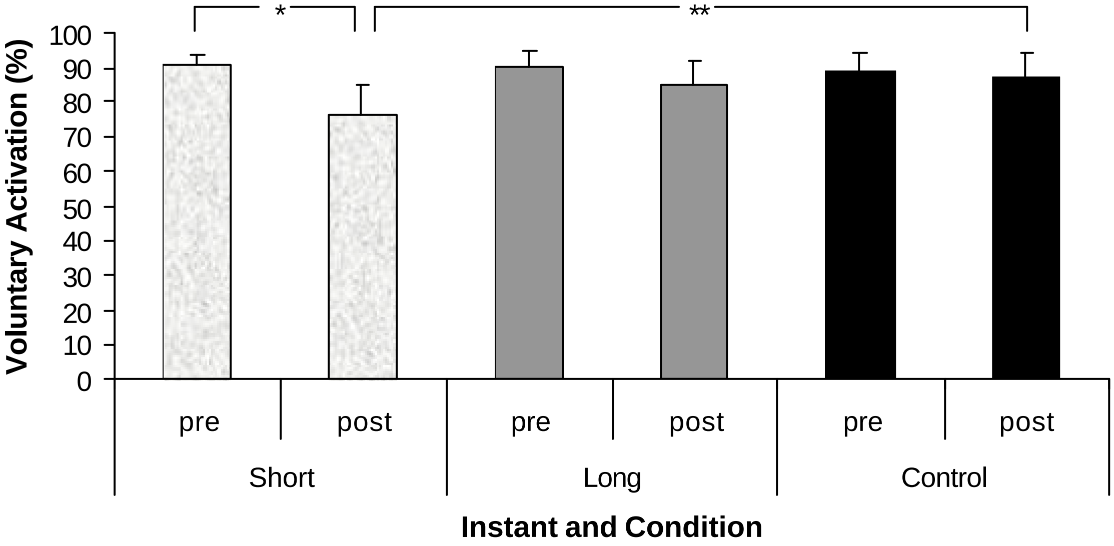


Figure 2.2. Mean and standard deviation of voluntary activation in different instants and conditions. *Difference between pre- and post- instant at short condition ($p < 0.01$); **Difference between short and control condition at post-instant ($p < 0.05$).

2.4. Discussion

The present study focused on neuromuscular responses after short- and long-term countermovement jump exercise to fatigue. After fatiguing protocols, $MVIC_E$ force decreased and it was accompanied by a reduced EMG activity, peak twitch torque and contraction time. In addition, the main difference between short- and long-term fatigue protocols was a voluntary activation deficit only after short-term exercising.

During muscle contraction, increased coactivation of the antagonist muscle can oppose the torque produced by the agonist muscle, decreasing net joint moment. In this study, muscle coactivation of BF did not change pre- to post-fatigue protocols and was therefore not responsible for the decrease in $MVIC_E$ after short- or long-term exercising.

Amplitude and duration of M-wave response indicated the stability of neuromuscular propagation throughout the muscle fibre. The stimulation response represents the conversion of an axonal action potential into a sarcolemmal action potential (Gandevia, 2001). In the present study, neither the amplitude nor the duration of M-wave response changed after either fatiguing protocols, suggesting that neuromuscular propagation was preserved. On the other hand, muscle contractile ability decreased. Peak twitch torque (short = -26%, long = -32%) and contraction time (short = -22%, long = -13%) decreased after both fatiguing protocols confirming the hypothesis that muscle tissue had a decreased ability to produce force.

After short- and long-term fatiguing exercise there were RMS_{VL} activation deficits (short = -22% and long = -14%). Nevertheless, there was significant voluntary activation deficit only after short-term SSC fatigue (-14%), detected by ITT. The validity of surface EMG measurements to detect decrements in neural drive with fatigue has been questioned because of signal repeatability, cancellation and synchronization (Day & Hulliger, 2001, Dimitrova & Dimitrov, 2003, Place *et al.*, 2007, Yao *et al.*, 2000). However, ITT has been used to estimate voluntary activation level (Bampouras *et al.*, 2006, Place *et al.*, 2007) which consists of applying a supra-maximal electrical stimuli during a MVIC$_E$ to induce additional torque increments (Merton, 1954). If the electrical stimuli evoke additional torque, then the neural drive during a MVIC$_E$ is assumed to be submaximal. Therefore, the deficit in voluntary activation seems to be more substantial in the short- than in the long-term protocol due to the concomitant decrement in ITT and in RMS$_{VL}$ observed in this condition.

These results may suggest that fatigue during the short-term SSC protocol was produced by both decrement in voluntary activation and failure in the contractile machinery. On the other hand, fatigue during long-term SSC protocol was caused mainly by a decreased muscle contractile capacity based on electrically stimulated parameters. These results are in contrast to the hypothesis based on the findings of previous studies and they may be explained by differences in the protocols, as explained further.

Impairment in the muscle contractile machinery has been found after SSC fatiguing exercise. In this study, peak twitch torque declined after short- and long-term fatigue, which is in accordance with other fatigue studies (Kuitunen *et al.*, 2004, Skurvydas *et al.*, 2000, Strojnik & Komi, 1998). The reduced twitch torque can be associated with a reduced Ca^{++} release and uptake from the sarcoplasmic reticulum and/or reduced capability of cross bridges to form strong binding (Duchateau & Hainaut, 1985, Metzger & Moss, 1990). Ca^{++} release and uptake seems to be dependent of exercise intensity (Ward *et al.*, 1998). However, there was no difference in MVIC$_E$ and PT drop between conditions, which may indicate a similar degree of contractile machinery failure. In addition, it has been suggested that a low muscle pH may impair force production and that blood lactate is an indirect marker of muscle pH (Fitts, 1994). However, Bogdanis and colleagues (Bogdanis *et al.*, 1995) found no relationship between muscle pH and blood lactate. Thus, even though blood lactate is an indicator of exercise intensity and blood lactate increased more after the short-term than the long-term jumping protocol no conclusion can be drawn about failure in actin-myosin binding. Therefore, these data may suggest that changes in Ca^{++} uptake and release were responsible for the drop in muscle force production in both fatiguing protocols.

After the short-term fatiguing protocol, reduced peak twitch torque was accompanied by a voluntary activation deficit. The mechanisms underlying voluntary activation deficit have been extensively discussed (Gandevia, 2001). After SSC fatiguing exercise, neural failure and/or adjustments may be expected to take place and to affect the activation pathways at different levels (Nicol *et al.*, 2006). The ITT has indicated central activation failure but it is not possible to identify exactly which mechanism (supraspinal, or disfacilitation of the a-motoneurone pool, or reflex inhibition) would be responsible for such changes. Reduced stretch-reflex sensitivity suggests the possibility of fatigue on the intrafusal fibres, which leads to a decreased excitability of the stretch reflex (Bongiovanni & Hagbarth, 1990). After short- (Nicol *et al.*, 1996) and long-term (Kuitunen *et al.*, 2004) SSC fatiguing, the stretch-reflex sensitivity has been reduced. A drop in the gamma loop activity may explain the decreased stretch reflex responsiveness and support of muscle spindle activity to alpha motoneurons (Hagbarth *et al.*, 1986). Further, reduced Hoffman reflex (H-reflex) sensitivity can indicate the excitability of the a motoneurone pool and it relies on the selective activation of group Ia afferent neuron (Gandevia, 2001). In addition, both reduced stretch- and H-reflex have been reported after short-term SSC fatiguing (Avela *et al.*, 2006). We also cannot rule out the possibility that muscle damage or fatigue may have activated groups III and IV afferents, which are able to produce autogenic inhibition and decrease the firing rate of alpha motoneurons (Bongiovanni & Hagbarth, 1990). Although stretch-, H-reflex, and muscle damage were not measured in our study, they are possible candidates for the decreased voluntary activation after the short-term fatiguing protocol.

Voluntary activation reduction has been found after long-term fatiguing (Kuitunen *et al.*, 2002, Kuitunen *et al.*, 2004), remaining unaltered (Avela *et al.*, 2006) or increased after short-term (Strojnik & Komi, 1998) fatiguing. The apparent disparity between the findings of the present and previous studies relating to reductions in activation level can be explained based on the nature of the exercise protocols. In contrast to the protocol adopted in the present study, where countermovement jumps were performed very close to maximal height (i.e., 95% of maximal height) and separated by a defined rest period, previous studies have performed submaximal (i.e., 70% of maximal height) rebound jumps continuously (Kuitunen *et al.*, 2002, Kuitunen *et al.*, 2004). A deficit in voluntary activation has previously been reported after prolonged continuous running (Millet *et al.*, 2002, Millet *et al.*, 2003) and cycling (Lepers *et al.*, 2002). The similarity among jumping, running and cycling fatigue studies is the voluntary activation deficit after performing continuous exercise. Evidence suggests that the development of central fatigue as exercise progress may vary according to the continuous/intermittent nature of the exercise task. This hypothesis was confirmed by showing an earlier and larger voluntary activation deficit during maximal continuous elbow isometric task than during the same task performed intermittently (Bilodeau, 2006). Bilodeau (2006) hypothesized that smaller changes in voluntary activation during an intermittent than continuous task could occur due to a rapid recovery of the changes in motor cortex excitability. However, after continuous and intermittent submaximal running, with no exhausting exercise, responses were independent of the running mode (Racinais *et al.*, 2007). Although jumps were always performed intermittently in our study, after short-term fatiguing protocol, the greater voluntary activation deficit may be due to the shorter resting interval between jumps. Shorter rest intervals may produce a high degree of fatigue in fast twitch fibers, which may produce a decreased muscle spindle activity and support to alpha motoneurons. This hypothesis is supported by the fact that high frequency tendon vibration has the ability to recover muscle force production capacity during maximal voluntary contractions in which fast twitch fibres are highly fatigued (Bongiovanni & Hagbarth, 1990). Then, in countermovement jumps in which fast twitch fibres are mainly evoked, the aforementioned hypothesis may support the voluntary activation differences between short- and long-term exercises.

2.5. Conclusion

The results indicated that shorter rest intervals between countermovement jumps produced both central and peripheral fatigue. On the other hand, longer rest intervals produced peripheral fatigue only. Such different causes of fatigue in countermovement jumps may influence on movement patter in a different way. Then, the next chapter investigates the influence of fatigue on movement pattern of countermovement jumps.

CHAPTER 3

The effects of fatigue on movement pattern of countermovement jumps

3.1. Introduction

Although there are many directions and types of mobility for human movements (degrees of freedom), they are performed in a controlled and a coordinated form (Bernstein, 1967). From this perspective of movement, Bernstein (1967) questioned how the motor system could control all motor actions through inputs sent to each motor unit of each muscle. A solution given to this issue was named coordinative structure (Kelso *et al.*, 1980, Kugler *et al.*, 1980). According to this solution, the motor system would constrain the functioning of many muscles leading them to act as a functional unit, instead of individually.

However, the performer (organism) is always moving in a specific context. Such a context is related to the environment and the task goal. Hence, the three aspects (organismic, environment, and task constraint) constrain the action (Newell, 1986). The interactions between these constraints lead to a specific control and coordination. Frequently, constraints are manipulated either through the environment or through the task, in order to determine the solution of motor system. The organism also has its own constraints, such as joint angle limit, muscle fatigue can also be understood as an organismic constraint (Rodacki et al. 2001, Rodacki et al. 2002). Thus, changes on the musculoskeletal system, due to fatigue, may lead to a new movement pattern in order to compensate for the force loss and, therefore, try to sustain the required external output (van Ingen Schenau *et al.*, 1995).

Studies have investigated the influence of fatigue on multi-joint tasks, for instance, lifting (Sparto *et al.*, 1997), throwing (Huffenus *et al.*, 2006), cycling (Hautier *et al.*, 2000), sawing (Cote *et al.*, 2002), and vertical jumping (Rodacki *et al.*, 2001). Usually, they have found changes in the timing and sequencing of both segmental movement and muscle activation (coordination variables), as well changes in the magnitude of the muscle activation and the force (control variables).

In vertical jumping, simulation (Bobbert & Van Soest, 1994) and experimental studies (Rodacki *et al.*, 2001) the importance of movement coordination on the performance has been shown. Vertical jumping performance is related to the ability not only to jump as high as an individual can, but also to sustain required target height. This one depends on delaying the process of fatigue (i.e., a reduction on maximal voluntary contraction), which can lead to the point of fatigue (i.e., inability to sustain required task goal). Usually, studies have compared the movement pattern performed before and after fatigue protocol. However, measurements throughout fatigue protocol are important, in order to determine whether adjustments in the movement pattern may occur during the process of fatigue (van Ingen Schenau *et al.*, 1995). Therefore, this study aimed to investigate the movement pattern of countermovement jump throughout fatigue exercise protocols. It was hypothesized that movement pattern of countermovement jump is (re)organised when it is performed under fatigue.

3.2. Methods

Participants. Seven healthy males volunteers (mean ± SD; 23.9 ± 1.8 years; 74.5 ± 7.7 kg; 1.78 ± 0.06 m), engaged in volleyball sport and with previous experience in vertical jump, participated in this study. Participants were asked to refrain from severe physical activity in the day before the experimental session. This study was approved by the research ethics committee of Sao Paulo State University and participants were informed about the procedures and risks before giving written consent.

Experimental Design. Participants performed seven experimental sessions across 30 days. Sessions were at least three days apart to avoid any fatigue carry over effect. The first experimental session was conducted to familiarize individuals to the study procedures. In this session, participants performed maximal countermovement jumps, followed by simulation test of intermittent jumping fatigue. After the familiarisation session, a set of three sessions (short-term sessions) were performed to estimate the critical interval (Pereira *et al.*, 2004), in which the participants performed countermovement jumps (CMJs) to a fixed height until volitional fatigue using different resting periods. Then, another set of three sessions (long-term sessions) were performed with three different resting intervals: (1) at critical interval; (2) 10% below critical interval; and (3) 10% above critical interval. The experimental design is schematized below (Figure 3.1).

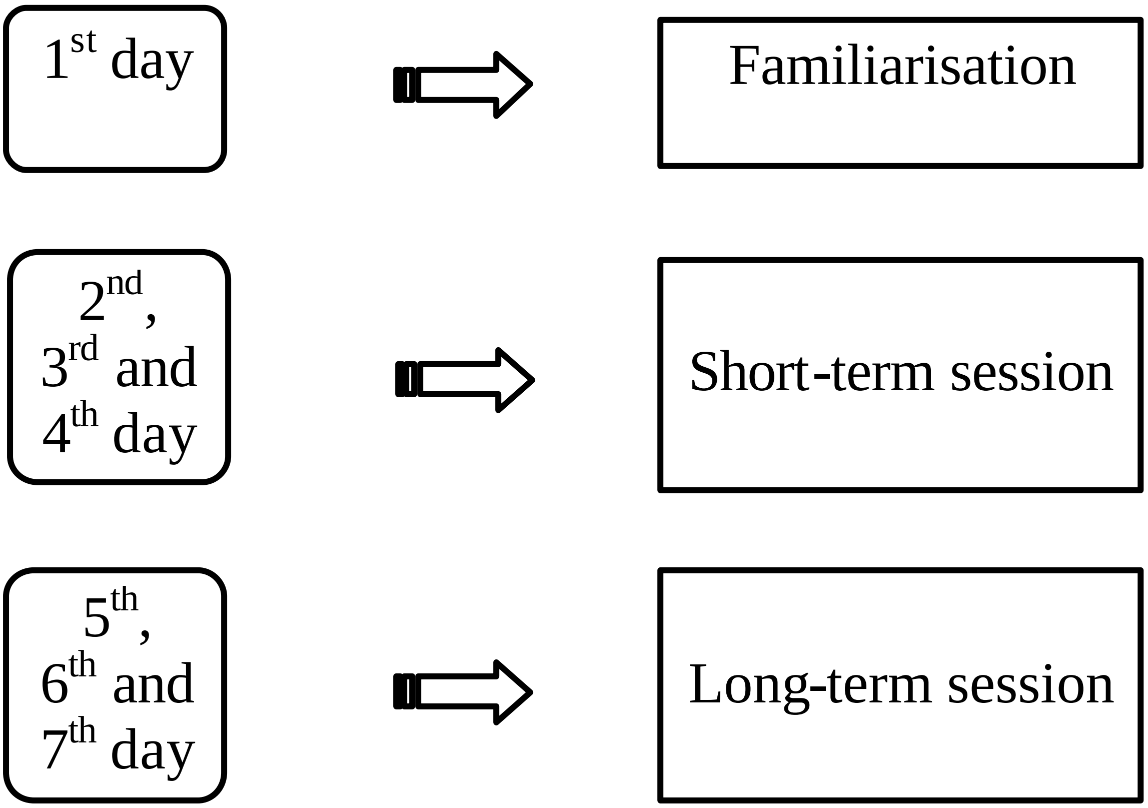


Figure 3.1. Schematic presentation of the experimental design.

Countermovement jumps. Before starting any session, participants were prepared for the electromyographic and kinematic assessments (see below), received explanation and demonstration of the countermovement jumps, and performed ten countermovement jumps (CMJs) to warm-up, which were followed by a set of five maximal CMJs. During the CMJs, participants bent their knees to a freely chosen angle, which was followed by a maximal vertical thrust. The effect of the arms was minimized by requesting the participants to keep their hands and forearms crossed against the chest. The average of five maximal height jumps was determined before each jumping fatigue protocol. The fatigue protocol consisted to the participants jumping at 95% of their maximal jump height, under different rest interval and until fatigue. Participants received visual feedback of their jump height immediately after each jumping, in which it was estimated by impulse-moment method (Linthorne, 2001). The point of fatigue was characterized as inability to sustain the target height for three consecutively jumps, despite verbal encouragement.

Kinematic assessment. Infrared emitters were placed on the right side of each participant's body to match with the following sites: 1) fifth metatarsal joint, 2) lateral malleolus, 3) lateral femoral epicondyle of the knee, 4) the most prominent protuberance of the greater trochanter, and 5) acromion process. The optoelectric system (Optotrak 3020 – 3D Motion Measurement System, NDI) was used to record (200 Hz) the x, y and z coordinates of the joint points and they were filtered using a zero lag fourth order low pass digital Butterworth filter. Cut-off frequency was defined by residual analysis and set at 7 Hz (Winter, 2005). The coordinates of the joint points were used to calculate joint angular displacement, velocity, and acceleration of the ankle, knee and hip. Figure 3.2 provides visual information of the four-segment model and also shows the joint angle conventions.

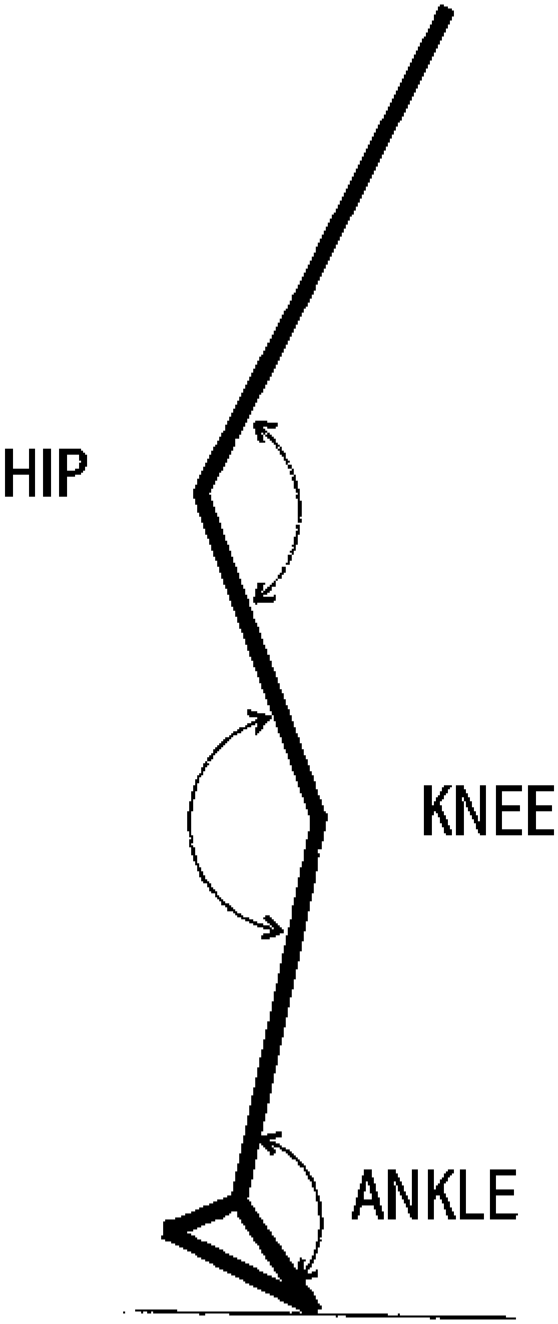


Figure 3.2. The four-body segment model and the joint angle convention (adapted from Rodacki *et al.*, 2001).

Electromyographic assessment. Surface electromyographic (EMG) signals were recorded from gastrocnemius medialis (GA), semitendinosus (ST), vastus lateralis (VL) and vastus medialis (VM) muscles, using four Ag-AgCl bipolar surface electrodes (Noraxon, USA), located according to the recommendations by SENIAM (SENIAM, 1999). Reference electrode was placed on the left wrist. Before placement of the electrodes, the skin was shaved, lightly rubbed with abrasive gel and cleaned with alcohol swabs to reduce interelectrode impedance below 5 KO. Due to the fast and explosive characteristic of the movement, all sites were covered with straps of adhesive tape to prevent disconnection and reduce movement artifacts. All test sites were identified, marked with indelible ink and prepared by the same experimenter. The EMG signals were differentially amplified with a gain of 1000 and a bandwidth of 10-500 Hz at –3 dB (EMG System, Brazil). Raw EMG activity was recorded with a sampling frequency of 1000 Hz and processed with a multichannel analog-to-digital converter (LabView, NI). The EMG data were rectified and filtered using a zero lag fourth order low pass digital Butterworth filter and cut-off frequency was set at 5 Hz.

Kinetic assessment. A force platform (Kistler, model 9286A, Winterthur, Switzerland) synchronized with the kinematic and electromyographic data measurements and sampling at 1000 Hz provided force-time traces. The impulse was calculated by integrating the force-time curves of the vertical component of the ground reaction forces during the positive phase of the movement (see below). The ground reaction forces were reduced to 200 Hz and combined with kinematic analysis in order to calculate net moments at the ankle, knee and hip joints. Extensor moments were considered positive at all joints. The moment of inertia of each segment was estimated by using the Drillis and Contini equation (Drillis & Contini, 1966). Net powers around the joints were also calculated by multiplication of the net moments and joint angular velocities.

Definition of work to rest ratio. The work to rest ratio was manipulated using different resting intervals between CMJs, whilst jump height was pre-established. The short-term fatigue protocol consisted of three sessions of jumping until ~2 min (high), ~5 min (intermediate), and ~10 min (low), in order to determine the critical interval (CI) (Pereira 2004). Then, participants jumped three more sessions (long-term session), in which exercise duration should be ~20 min at critical interval intensity (at_{CI}), less than 20 min ($above_{CI}$), and greater than 20 min ($below_{CI}$).

Definition of variables. The movement duration was defined from the beginning to the take-off instants. Such movement duration was fractioned in three phases: the negative phase (NEG), the transient phase, i.e., the knee joint reversal (JR) from eccentric to concentric phase, and, the positive phase (POS). The initiation of negative phase (beginning of movement) was defined as the instant when the vertical force decreased by 5% of body weight. The end of the eccentric phase, that corresponds to the beginning of concentric phase, was defined as the instant in which the net impulse on unweighting phase was equal to the net impulse over body weight (Bosco & Komi, 1979). The end of positive phase was determined when the participants lost their contact with force platform (take-off) and the vertical component of the ground reaction forces was zero. The joint reversal time was determined as the period in which the angular velocity ranged between +30 $deg.s^{-1}$ and -30 $deg.s^{-1}$ in relation to the deepest knee flexion angle (Rodacki *et al.*, 2001). The initiation of joint extension (IEX), the peak angular velocity

(PAV), the peak power of the ankle, knee and hip joints were determined. The IEX was defined as the first frame after a joint reaches its deepest flexion angle. The PAV and peak power were defined as the instant in which the greatest joint angular velocity and power around the joint, respectively, were achieved during POS phase. In addition, at IEX, PAV and peak power instants achievement were determined according to their time difference with take-off instant. These instants were normalized by jumping duration (NEG phase added to POS phase). The time difference in which IEX of each joint occurred was used to establish the sequential relationship between adjacent segments. If the movement of a proximal segment precedes the movement of its distal counterpart (i.e., a proximal-to-distal order), the time difference, between the IEX, will be negative. The EMG traces were normalized according to the signal magnitude. For each muscle, the greatest electromyographic signal value obtained during the performance of five maximal CMJs, performed before starting each session, was used as reference and set at 100%. Muscle activation was quantified during the POS phase of the movement by dividing the integrated EMG signal (the area under muscle activation curve) by the duration of the movement phase. The timing of each muscle's initiation of activation (IACT) was also examined. IACT was arbitrarily considered as the instant in which the EMG traces were equal to 20% of the highest muscle activation obtained during the POS phase (Rodacki *et al.*, 2002). This instant was subtracted from take-off instant and normalized by jumping duration. The sequential orders of the muscle initiation activations were also analyzed subtracting their IACTs. Knee joint stiffness was determined by calculating the coefficient of linear regression of the moment-angle relationship from the last 15 degrees of the NEG phase to the deepest knee joint excursion (Rodacki *et al.*, 2002). For all data analyses and determination of the variables, a routine was made using Matlab program (Math Works, Inc., Natick, Massachusetts, USA). See Appendix A to details of routine in Matlab.

Statistical analysis. For each intensity and participant, thirty CMJs were considered to be analyzed, despite the total number of movements performed during the session. The average of the first ten jumps, the ten jumps in middle, and the last ten jumps of each session were analyzed. These jumps were divided into four periods: from the 1st to the 10th jump (I0), from the 11th to the 20th jump (I50), from the 21st to the 27th jump (I90), and from the 28th to the 30th jump (I100). After normality testing, differences in all variables between test conditions were tested with mixed models (SAS 8.2, SAS Institute Inc., North Carolina, USA). Work to rest ratio, varying according to different rest period (high, intermediate, low, above$_{CI}$, at$_{CI}$, and below$_{CI}$), and period (I0, I50, I90, I100) were considered as fixed factors and participant as a random factor. For number of countermovement jumps, exercise duration and rest interval, work to rest ratio was considered as fixed factor and participant as random factor. A post-hoc test, with an adjustment by Tukey, was used for multi-comparison purposes. Significance was set at $p < 0.05$. Data are presented as mean ± SD.

3.3 Results

Countermovement jump performance. Different rest period (i.e., work to rest ratio) between CMJs in this study was proven to successfully promote different number of CMJs and, consequently, different time to fatigue (Table 3.1). The jump height decreased significantly at I90 (-3%; p<0.01) and at I100 periods (-6%; p<0.01) compared to both I0 and I50 (Table 3.2). In addition, at I100 period the jump height decreased 3% compared to I90 (p<0.01). However, there was no significant interaction between work to rest ratios and periods for any variables measured. There was no significant work to rest ratio main effect, but there was significant Period main effect. Thereby, all results are presented regarding different periods (I0, I50, I90 and I100).

Table 3.1. Number of countermovement jumps, exercise duration and rest interval performed during six different sessions.

	N° of CMJs	Exercise Duration (s)	Rest Interval (s)
High	31 ± 10	157.7 ± 56.8	5.0 ± 0.8
Intermediate	59 ± 14	355.1 ± 53.2	6.2 ± 1.2
Low	84 ± 30 a	596.7 ± 136.3 a	7.4 ± 1.6
Above$_{CI}$	109 ± 62 a	941.3 ± 351.4 a, b	9.3 ± 3.0 a, b
At$_{CI}$	114 ± 34 a, b	1110.7 ± 247.8 a, b, c	10.3 ± 3.1 a, b, c
Below$_{CI}$	138 ± 61 a, b, c	1443.2 ± 373.1 a, b, c, d	11.4 ± 3.6 a, b, c

aSignificant different from high (p<0.01); bSignificant different from intermediate (p<0.01); cSignifcant different from low (p<0.01); dSignficant different from above$_{CI}$ (p<0.01).

Kinetics and Kinematics. For I90 and I100 periods, the NEG phase duration was 3% and 5% ($p<0.01$) greater than I0 (Table 3.2). From I50 to I100 periods, NEG phase duration increased 4% ($p<0.01$). The POS phase duration increased 5% ($p<0.01$) from I0 to I50 period, after that, similar values were maintained towards the end of the session (Table 3.2). The JR did not change significantly ($p>0.05$) throughout the CMJs. During POS phase, the impulse reduced 1% ($p<0.05$) from I50 to I90 period. The concentric impulse also reduced 2% ($p<0.01$) at I100 compared to I0 and I50 period and, it reduced 1% ($p<0.01$) from I90 to I100 (Table 3.2). The vertical ground reaction force traces of a participant are presented on figure 3.6.

Neither IEX ankle joint angle nor its time changed throughout the periods ($p>0.05$). Nevertheless, IEX knee joint angle at I0 decreased at I50 ($p<0.01$) and it was maintained throughout the remaining time (Table 3.2 and Figure 3.3), although there was no statistical significance comparing to I0. Related to this response, knee joint stiffness reduced 10% ($p>0.01$) only at I50 compared to I0, and it re-established initial values throughout the remaining jumps. The time of IEX knee joint started earlier at I50 (7%; $p<0.01$) and at I90 (10%; $p<0.01$) compared to I0 instant. It also started 4% (I100) earlier than at I50. The IEX hip joint angle at I50, I90 and I100 decreased 6% ($p<0.01$) compared to I0 (Table 3.2 and Figure 3.3). The time of IEX hip joint followed the same pattern of knee joint, which started earlier at I50 (5%; $p<0.01$) and at I90 (6%; $p<0.01$) compared to I0. In addition, at I100, IEX hip started 3% ($p<0.01$) earlier than at I50. The time of PAV ankle joint started 18% ($p<0.01$) at I100 earlier than at I0. The time of peak power ankle joint started earlier 2%, 3%, and 4% ($p<0.01$) at I50, I90 and I100, respectively, compared to I0. However, when the time of IEX, PAV and peak power were normalized by jumping duration, there was no significant difference for any joint. The delays between IEX_{HIP}-IEX_{KNEE} and IEX_{KNEE}-IEX_{ANKLE} indicated the existence of a proximal-to-distal order, in which the hip was consistently the first joint to extend and was followed by knee and ankle joint extensions (Table 3.2).

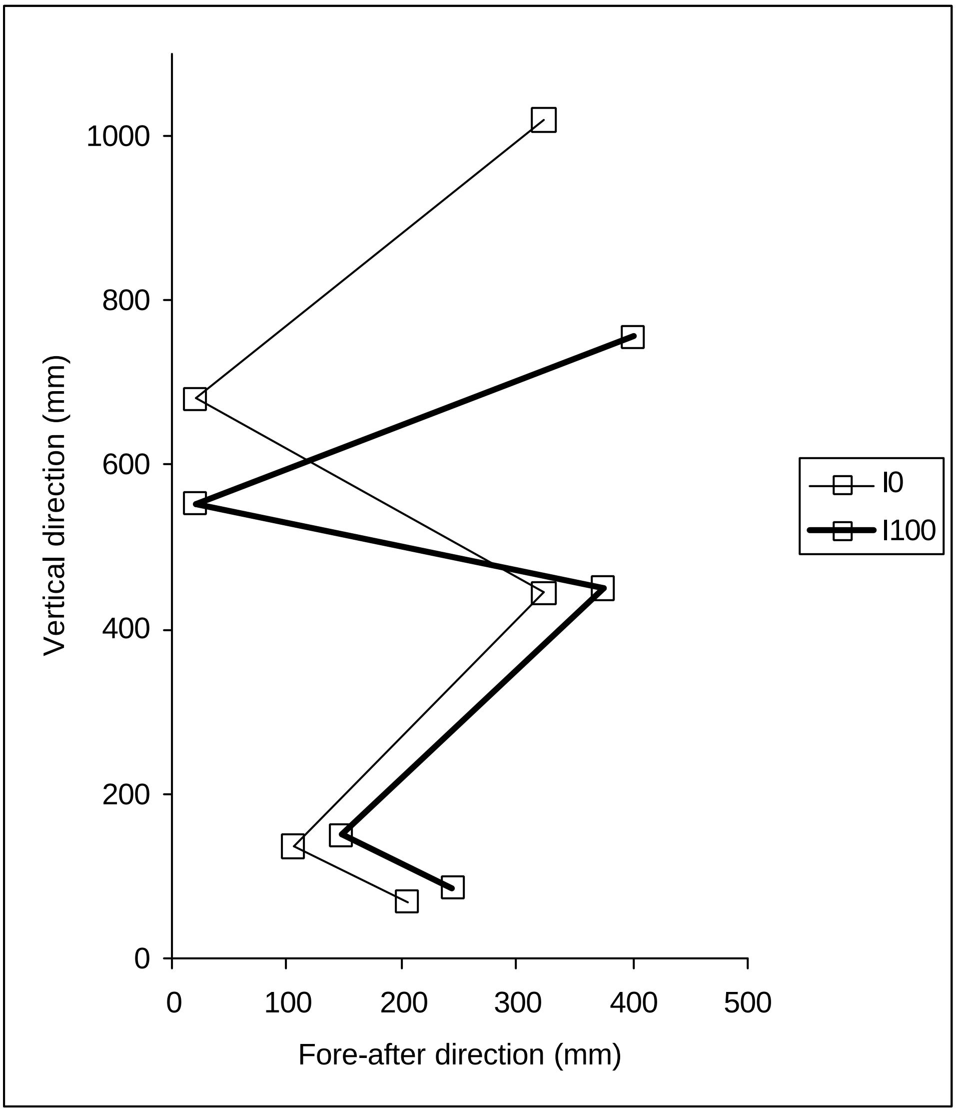


Figure 3.3. The four-body segment model at I0 (non-fatigued) and at I100 (fatigued) period of a participant.

Table 3.2. Kinematic and kinetic characteristics of countermovement jumps performed at four different periods.

	PERIODS			
	I0	I50	I90	I100
Jump Height (m)	0.32 ± 0.06	0.32 ± 0.06	0.31 ± 0.06 a,b	0.30 ± 0.06 a,b,c
NEG duration (ms)	519 ± 39	525 ± 0.054	536 ± 59 a	544 ± 6 a,b
JR duration (ms)	145 ± 46	151 ± 52	164 ± 55	194 ± 184
POS duration (ms)	319 ± 21	334 ± 0.025 a	339 ± 25 a	342 ± 29 a
Impulse (N·s)	228.3 ± 21.2	228.2 ± 21.6	226.0 ± 22.0b	224.5 ± 29.0 a,b,c
Knee Stiffness ($N\cdot m.deg^{-1}$)	106.6 ± 32.5	96.4 ± 35.8 a	103.6 ± 40.9	104.1 ± 37.2
ANKLE				
IEX joint angle (deg)	88.2 ± 5.5	87.6 ± 5.3	87.6 ± 5.7	87.7 ± 6.0
IEX time (ms)	-465 ± 124	-483 ± 137	-494 ± 146	-496 ± 163
PAV, POS ($rad.s^{-1}$)	14.1 ± 2.8	14.1 ± 2.7	14.1 ± 2.5	14.1 ± 2.8
PAV time, POS (ms)	-40 ± 16	-42 ± 17	-44 ± 17	-47 ± 18 a
Peak Power, POS (W)	1,317.9 ± 77.4	1,314.7 ± 79.2	1,315.6 ± 88.2	1,317.6 ± 91.7
Peak Power, time (ms)	-130 ± 11	-133 ± 12 a	-134 ± 11 a	-135 ± 12 a
KNEE				
IEX joint angle (deg)	76.6 ± 9.6	74.1 ± 9.3a	74.6 ± 9.7	73.5 ± 10.2
IEX time (ms)	-505 ± 75	-541 ± 79 a	-555 ± 80 a	-563 ± 87 a,b
PAV, POS ($rad.s^{-1}$)	15.4 ± 2.2	15.4 ± 3.2	15.4 ± 2.2	15.4 ± 3.2
PAV time, POS (ms)	-58 ± 16	-57 ± 18	-56 ± 18	-59 ± 19
Peak Power, POS (W)	1,708.7 ± 203.0	1,720.1 ± 200.3	1,702.7 ± 190.3	1,718.6 ± 192.0
Peak Power, time (ms)	-182 ± 16	-183 ± 16	-183 ± 18	-186 ± 22
HIP				
IEX joint angle (deg)	69.5 ± 9.0	65.2 ± 9.0a	65.2 ± 8.6 a	65.4 ± 9.4 a
IEX time (ms)	-588 ± 30	-615 ± 34 a	-624 ± 38 a	-632 ± 41 a,b
PAV, POS ($rad.s^{-1}$)	14.3 ± 2.3	14.3 ± 3.3	14.2 ± 2.3	14.2 ± 1.3
PAV time, POS (ms)	-66 ± 17	-65 ± 19	-65 ± 21	-67 ± 22
Peak Power, POS (W)	1,556.2 ± 142.2	1,543.5 ± 142.2	1,533.8 ± 143.2	1,521.0 ± 157.8
Peak Power, time (ms)	-248 ± 70	-264 ± 79	-259 ± 68	-269 ± 90
Delay between IEX_{HIP}-IEX_{KNEE} (ms)	-83 ± 59	-75 ± 60	-69 ± 60	-69 ± 61
Delay between IEX_{KNEE}-IEX_{ANKLE} (ms)	-40 ± 119	-57 ± 131	-61 ± 138	-67 ± 153

NEG: negative phase; JR: knee joint reversal; POS: positive phase; IEX: initiation of the joint extension; PAV: peak angular velocity. aSignificant different from I0; bSignifcant different from I50; cSignificant different from I90.

Electromyographic analysis. The EMG of GA muscle decreased significantly (p<0.01) from I0 to I50 (-12%), to I90 (-20%), and to I100 (-20%) periods (Figure 3.4). In addition, GA muscle activation at I90 and at I100 reduced significantly compared to I50 (p<0.01). There was a significant reduction (-10%; p<0.01) on VM muscle activation at I90 and at I100 compared to I0. The ST and VL muscles activation did not change (p>0.05) throughout the periods (Figure 3.4). The timing and sequencing of muscles activation did not change throughout the exercise (Figure 3.5). The electromyographic traces of a participant are presented on figures 3.7 and 3.8.

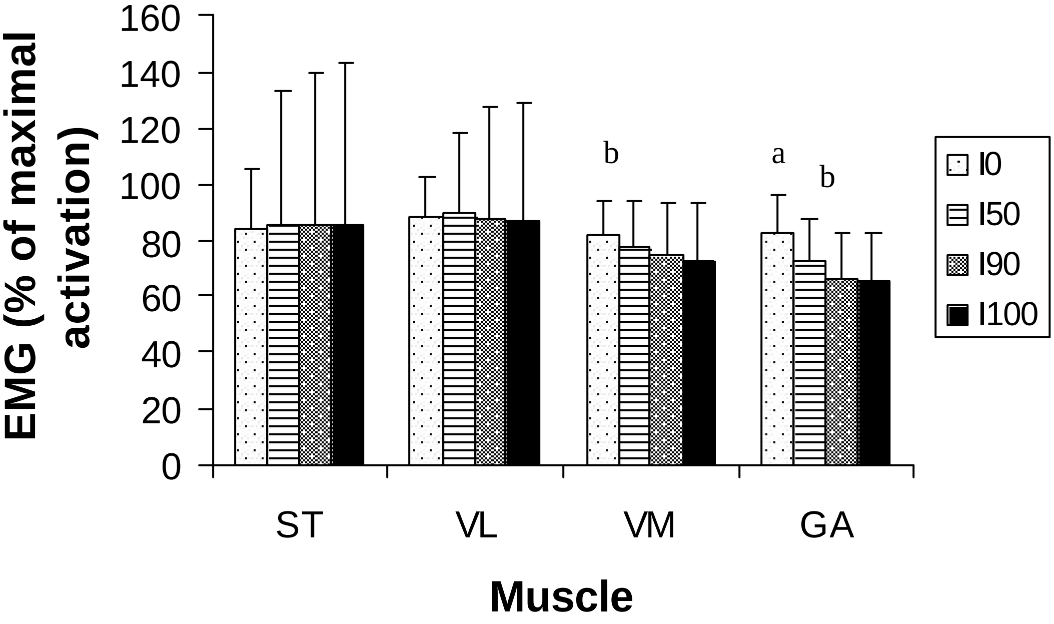


Figure 3.4: Mean and standard deviation of electromyographic activity (EMG) of semitendinosus (ST), vastus lateralis (VL), vastus medialis (VM), and gastrocnemius (GA) muscles during four different periods. [a]Significant different from I50, I90 and I100 ($p<0.01$); [b]Significant different from I90 and I100 ($p<0.01$).

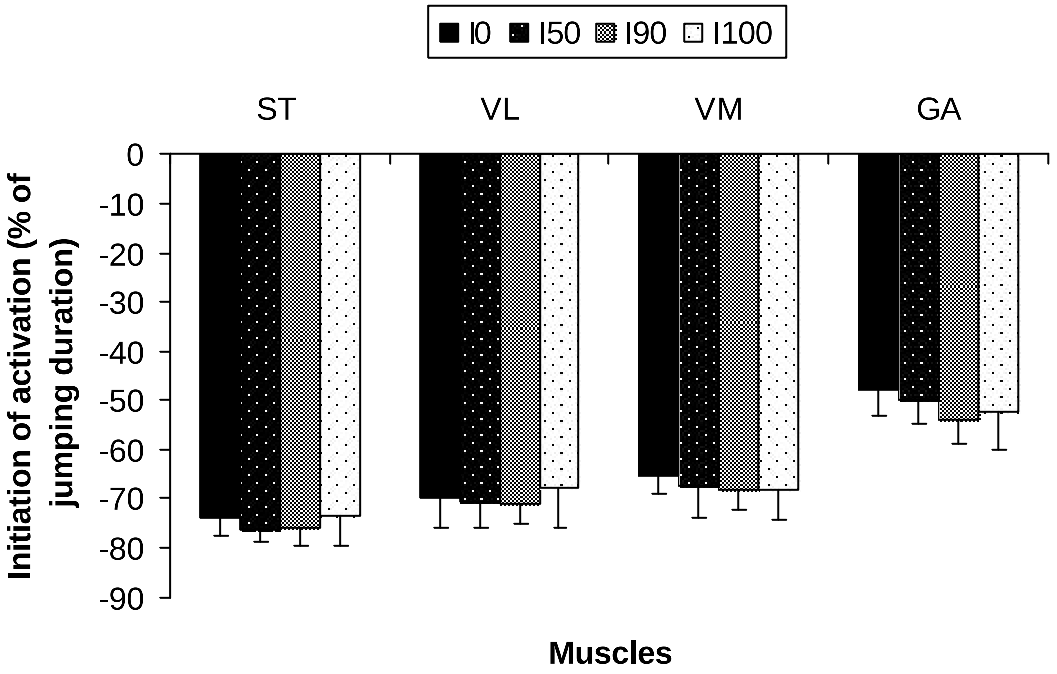


Figure 3.5: Mean and standard deviation of time difference between the initiation muscle activation and take-off instant of semitendinosus (ST), vastus lateralis (VL), vastus medialis (VM), and gastrocnemius (GA) muscles during four different periods.

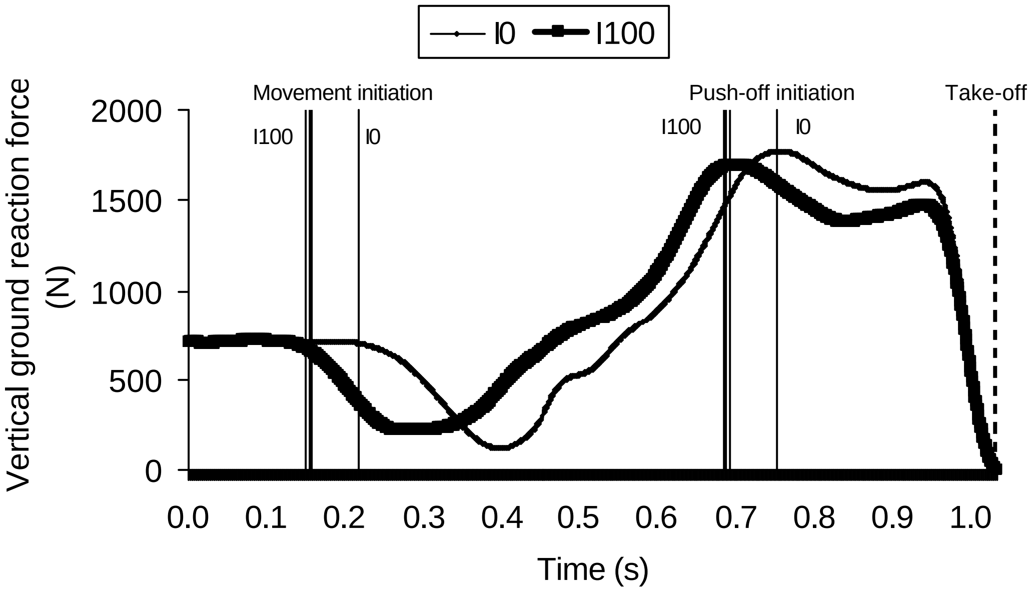
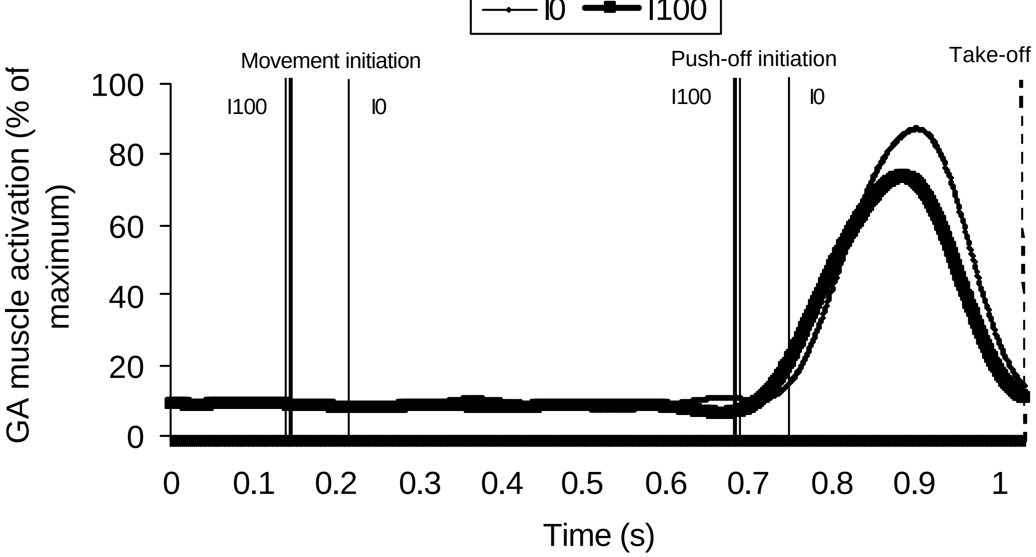


Figure 3.6: Vertical ground reaction force traces during two countermovement jumps performed at I0 (nonfatigued) and at I100 (point of fatigue) periods of a participant. Jumping phases are characterized as eccentric (from movement initiation to push-off initiation) and concentric phases (from push-off initiation to take-off instant).



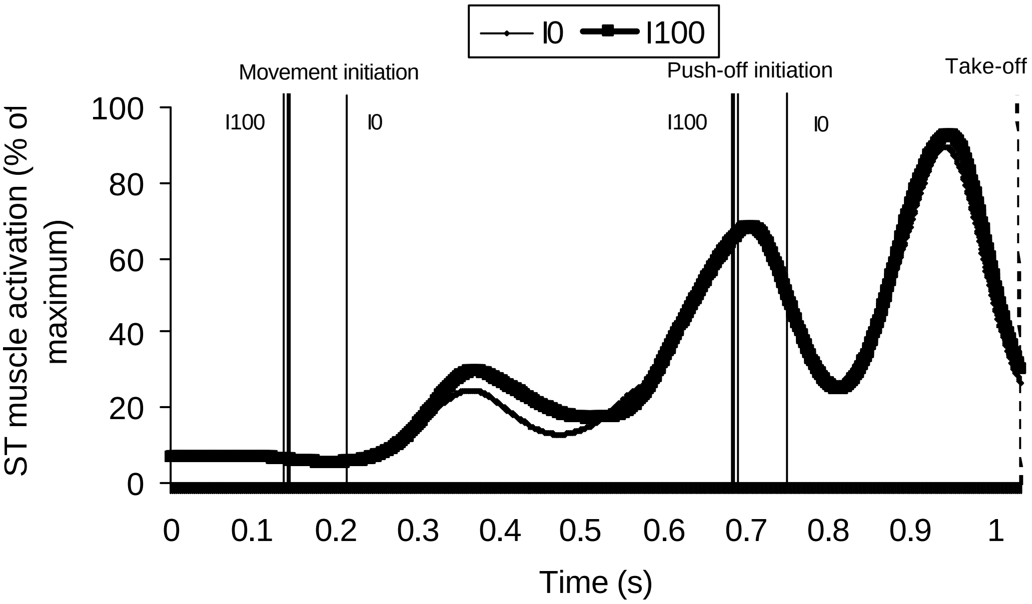


Figure 3.7: The electromyographic signals of gastrocnemius (GA) and semitendinosus (ST) muscles during two countermovement jumps performed at I0 (nonfatigued) and at I100 (point of fatigue) periods of a participant. The muscle activation was normalized with respect to maximal activation during maximal jumping. Jumping phases are characterized as eccentric (from movement initiation to push-off initiation) and concentric phases (from push-off initiation to take-off instant).

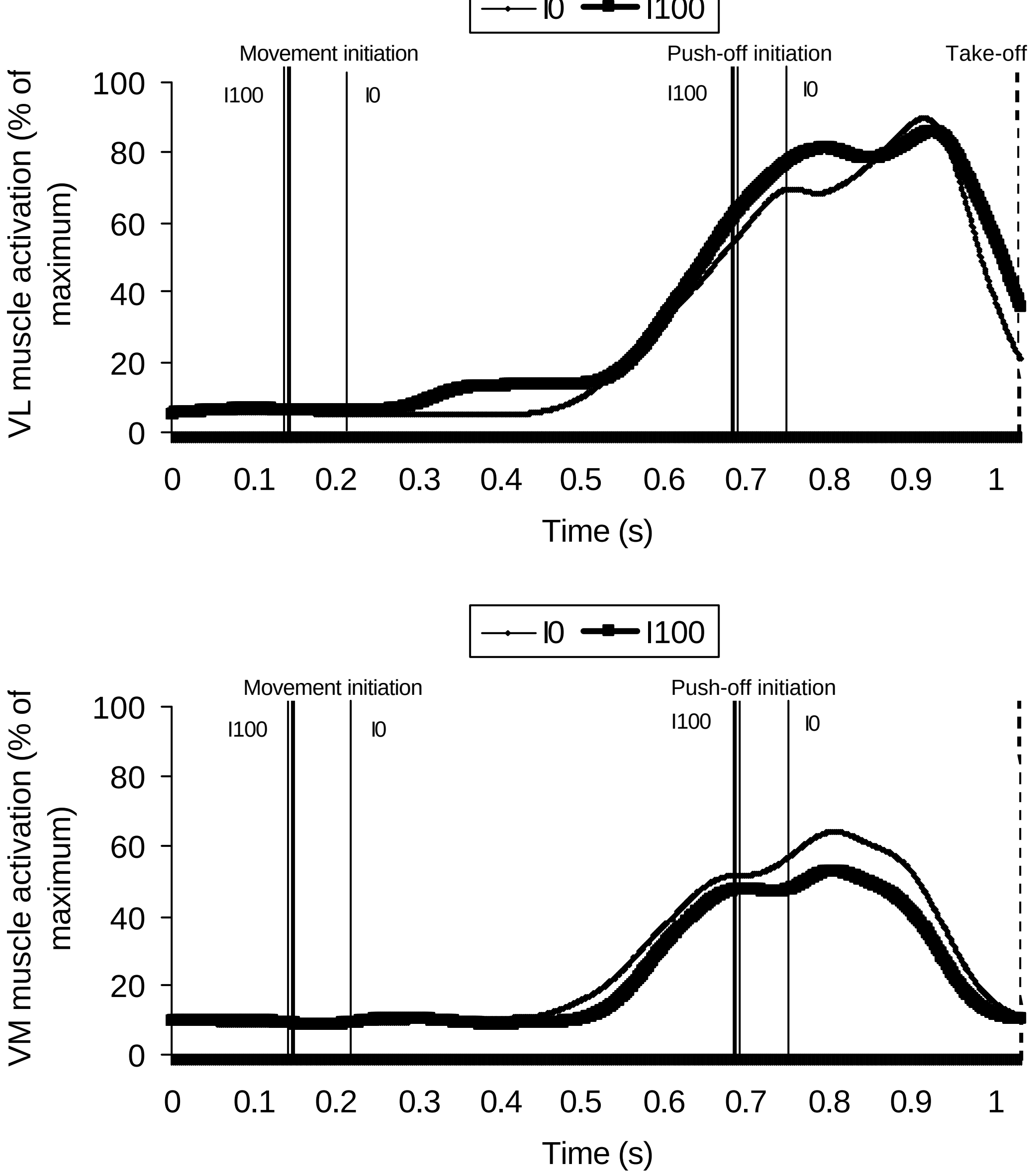


Figure 3.8: The electromyographic signals of vastus lateralis (VL) and vastus medialis (VM) muscles during two countermovement jumps performed at I0 (nonfatigued) and at I100 (point of fatigue) periods of a participant. The muscle activation was normalized with respect to maximal activation during maximal jumping. Jumping phases are characterized as eccentric (from movement initiation to push-off initiation) and concentric phases (from push-off initiation to take-off instant).

3.4. Discussion

This study aimed to investigate the control and the coordination patterns of countermovement jump throughout fatigue-induced exercise protocols. The results indicated that timing and sequencing of inter-segmental movement and of muscle activation remained relatively unaltered under fatigue. However, a few adjustments were performed on control patterns in fatigued jumps, in order to sustain the required jump height. These results were similar irrespective of fatigue-induced exercise work to rest ratio.

In vertical jumping, how ankle, knee and hip joints move to propel vertically the centre or mass is very important. The changes at these joints at the reversal point, which influences the coordination pattern of movement, cannot change without the application of some force, presumably a muscular force. Then, it can be expected that segmental movement coordination is a consequence of the muscle activation pattern, in which signals are coming from central nervous system. According to this assumption, changes in muscle activation pattern may lead to changes in segmental movement. Considering that, the argument proposed by van Ingen Schenau *et al.* (1995) that alterations in the musculoskeletal system, due to fatigue, may lead to changes in muscle activation timing and sequencing, in order to sustain the required task goal was not confirmed. In the present study, timing and sequencing of muscle activation were similar throughout the fatigue protocols. This can explain the similarities found in the topological characteristics of the movement (used to describe segment motions relative to each other) and the consistent pattern employed throughout fatigued jumps. Such a topological characteristic is the proximal-distal order (hip-knee-ankle joint extension) of segment motion maintained throughout fatigue protocols. A similar movement pattern in terms of muscle activation and segmental movement order was also observed on the literature (Bobbert & van Ingen Schenau, 1988). The idea that the motor system is constrained to function as a coordinative structure is an attractive way to explain this coordinated muscle activation and body motion pattern, irrespective of fatigued or nonfatigued condition. Movement pattern is a reflection of self-organising optimality of the biological system (Newell, 1986), in which the many degrees of freedom are reduced (or frozen) consistently and few adjustments are performed by the sub-systems according to the needed.

Increased contact time (eccentric added to concentric phase), before take-off instant, can be understood as a compensatory mechanism (adjustment) to sustain the required target height (Hortobagyi *et al.*, 1991). The concentric impulse is closely related to vertical velocity of mass centre at take-off instant, and this is related to the height achieved by centre of mass during flight phase. Therefore, the maintenance of jump height depends on, among others, the maintenance of concentric impulse. Since great amplitude of vertical ground reaction force has not been sustained under fatigue, longer movement duration (i.e., increased contact time) is able to sustain similar concentric impulse (Figure 3.6). The increased contact time may be as a result of larger range of motion at the knee and hip joints and/or increased joint reversal time (Hortobagyi *et al.*, 1991). As the joint reversal time did not change throughout the fatigue protocols and, knee and hip joint flexions were increased (Figure 3.3), these actions can explain increased contact time. Further, the muscle contractile impairment is frequently reported in exercises involving stretch-shortening cycle (Skurvydas *et al.*, 2000, Strojnik & Komi, 2000), which was also reported after fatigue using exercise protocol similar to this chapter (see details on chapter 2). Increased eccentric phase duration can be due to individuals start to resist the first part of the descending phase of the movement slightly earlier, i.e., before the knee joint reaches a deep excursion into a flexed position. This may be a strategy used to avoid muscle damage and improve the control of the movement in a situation where the participants do not rely on the ability of their fatigued muscles to react promptly and control the descending phase (Rodacki *et al.*, 2001). The muscle contractile impairment can also provoke a reduction on rate of twitch torque development, in such a way that may decrease the force development during the concentric phase of vertical jumping. Hereby, impairments on muscle contractile properties should have influenced the increased contact time during the vertical jumping.

The increased contact time duration is associated with other adjustments on control of movement observed throughout the exercise (starting from I50 period), and not only at the point of fatigue (I100 period). The alterations of control can be pointed by increasing knee and hip joint flexions, and by decreasing amplitude of GA and VM muscles activation. Increased joint flexions may lead to improve the net joint contributions in the total work done, in order to compensate the muscle force loss. Considering muscle activation amplitude, reduced activity of GA muscle can be explained due to its relatively fast-twitch characteristic. After maximal hopping fatigue, decreased activity of GA muscle before slow-twitch soleus muscle was found, suggesting evidence of selective fatigue (Moritani *et al.*, 1990). On the other hand, VM muscle activity reduced under fatiguing, whilst VL muscle activity did not change. In the present study, the motor task (countermovement jump) may influence on these muscle activation magnitudes, although VL muscle activation data is more reliable than VM muscle and it is considered a representative muscle of the quadriceps during knee extension and leg press (Alkner *et al.*, 2000).

Studies have demonstrated earlier achievement of peak angular velocity, peak power, and initiation of extension in vertical jumping, sawing and throwing performed under fatigue (Cote *et al.*, 2002, Huffenus *et al.*, 2006, Rodacki *et al.*, 2001). It means that the coordination of movement is reorganized under fatigue. In the present study, similar changes were observed. However, these earlier achievements were not supported when the timing was normalized by jumping duration. Fatigue has been compensated by a new motor coordination assuming new trade-offs between inter-segmental organization during hopping and throwing motor tasks (Bonnard *et al.*, 1994, Forestier & Nougier, 1998). It is necessary to bear in mind the nature of motor task before doing comparisons. For instance, it can be speculated that a hopping task does not involve all major muscles of the lower limbs and it increases the number of degrees of freedom, thereby individuals can compensate for the effects of fatigue and sustain the jump height. Related to throwing a ball, such motor task goal requires movement precision with low level of muscle strength. Both tasks are quite different from vertical jumping, which involves many joints and muscles, and high level of muscle strength. Such characteristics of maximal vertical jump decrease the possibility for the motor system to emerge a new movement pattern, due to the motor task constrains itself. Therefore muscle fatigue did not result in a new movement pattern (reorganisation) of the motor system.

The adjustments on control patterns may be considered as a compensatory mechanism in order to sustain the task goal, whilst a consistent signalisation from the neural central system was held to perform a similar movement pattern throughout exercise. In general, the movement pattern is the output of integration among environment, task and organism (Newell, 1986). In the present study, environment and task goal were maintained throughout the exercise and fatigue was the time dependent constraint which caused adjustments in control of movement. Interestingly, even at fatigue point, the motor system used similar coordination since I50 instant. It means that great changes on movement pattern may not occur due to the requirements that motor task imposed on the neuromuscular system, and small changes may occur at the beginning of the fatigue protocol to promote small adjustments in how the movement is performed in order to sustain the required task goal.

3.5. Conclusion

A new movement pattern of countermovement jump did not emerge under muscular fatigue. However, adjustments in control of movement are performed in order to sustain the required target height. These results suggest that the organisation of the movement pattern occurs since the beginning of exercising and the sub-systems - through integration among environment, task and organism - should adjust the movement to achieve the task goal.

CHAPTER 4

General discussion, conclusions and recommendations for future studies

4.1 General discussion

The present thesis aimed to explore the causes and consequences of fatigue in countermovement jump performed under different work to rest ratio. To answer these aims, two studies were performed in which the first one, described in chapter 2, focused on the following experimental aim:

1. To compare the causes of fatigue development during a short- and a long-term countermovement jump protocol through the manipulation of resting interval.

The research hypothesis to the first experimental aim was:

1. The cause of fatigue in countermovement jump is different when it is performed under different intensities.

In fact, short- and long-term fatigue-inducing exercise protocols presented different causes of fatigue after countermovement jumps. After the short-term fatigue protocol, maximal voluntary isometric contraction reduction was accompanied by reduced voluntary activation level (central fatigue) and peak twitch torque (peripheral fatigue). On the other hand, after the long-term fatigue protocol, maximal voluntary isometric contraction reduction was accompanied by reduced peak twitch torque only.

Although the first research hypothesis was accepted, the results contradict with the literature. It seems that short-term SSC exercises induce peripheral fatigue, whilst long-term SSC exercises induce central fatigue. However, even with this apparent pre-determined cause of fatigue in short- and long-term SSC-type fatigue, the results can be contrary due to the features of exercise employed to induce fatigue. For instance, both short- and long-term SSC fatigue protocols can induce central fatigue either at spinal or at supraspinal level (Kuitunen *et al.*, 2002, Nicol *et al.*, 1996). Interestingly, many studies that performed short-term SSC exercise reported fatigue at spinal level (Avela *et al.*, 2006, Horita *et al.*, 1996, Nicol *et al.*, 1996). Unfortunately, in the experimental study addressed to cause of fatigue, there was no measurement to test the integrity at spinal level, such as short-latency reflex. This is a limitation of present study and yet the results cannot rule on spinal or supraspinal influence on central fatigue.

Despite this limitation, the degree of central fatigue after short-term exercise was clearly different from long-term exercise. In the present study, small differences (~2 s) in resting interval between jumps determined different causes of fatigue. The task-dependency concept may explain such differences. This theme embodies the concept that fatigue is not the consequence of a single mechanism or site but, rather, that it can be induced by a variety of mechanisms and/or sites (Enoka & Stuart, 1992). It occurs when detail(s) of the task vary and mechanism(s) underlying fatigue also vary. Furthermore, when the measurement is performed (e.g., immediately, 30 min, 2 h, 1-4 days after SSC exercise fatigue) influencing from neural compensations of the contractile failure to protective inhibitions of the fatigued muscles (Nicol *et al.*, 2006). Therefore, a plethora of details can change the cause of fatigue from one site to another. Thereby, in the present study, the task-dependency concept is an appealing explanation to the different causes of fatigue in countermovement jumps performed under different work to rest ratio.

Since the exercise protocol used on chapter 2 was able to induce muscle fatigue, it was used as an organismic constraint to investigate its effect on the movement pattern of countermovement jump. Then, the chapter 3 addressed on the following experimental aim:

2. To investigate the movement pattern of countermovement jump throughout fatigue exercise protocols.

The research hypothesis to the second experimental aim was:

2. The movement pattern of countermovement jump is (re)organised when it is performed under fatigue.

The movement pattern of countermovement jump was not reorganised since the timing and sequencing of inter-segmental movement and of muscle activation remained relatively unaltered under fatigue. Therefore, the second research hypothesis was not accepted. One possible explanation for this result is that the fatigue-induced degree of the participants was not enough to change the movement pattern of countermovement jump. In the present study, the point of fatigue was characterised when participants did not reach the targeted jump height fixed at 95%. Such inability to sustain required target height was accompanied by reduction on maximal voluntary isometric contraction of the knee extensor of 16% and 5% after short- and long-term exercise, respectively. Rodacki *et al.* (2001) fixed the point of fatigue when individuals did not achieve 70% of their maximal jump height, and movement pattern did not change. In another study, the fatigue protocol reduced 14% the peak torque of the knee extensor (Rodacki *et al.* 2002) and movement pattern did not change, as well. However, it was demonstrated clear modifications of the running pattern after a marathon only when participants achieved reduction in the knee extensor maximal voluntary contraction greater than 40% (Nicol *et al.*, 1991). Therefore, in the present study, the constraint imposed to the neuromuscular system was not enough to induce changes in coordinative structure in such a way that the movement pattern of countermovement jumping was reorganized.

From the energetic point of view to perform vertical jump, it was proposed that participants tried to achieve the submaximal targeted jump heights with minimal energy expenditure (Vanrenterghem *et al.*, 2004). Using simulated jumps with a forward dynamic musculo-skeletal model, this hypothesis was reinforced (Vanrenterghem *et al.*, 2006). The results demonstrated that as the target height was lowered, two major changes occurred in the motion pattern. First, the countermovement amplitude was reduced; this helped to save energy because of reduced dissipation and regeneration of energy in the contractile elements. Second, the contribution of rotation of the heavy proximal segments of the lower limbs to the vertical velocity of the centre of mass at take-off decreased. Assuming that movement pattern of submaximal vertical jump is changed in order to have minimal energy expenditure; it is likely that the minimal energy expenditure hypothesis remains under fatigue or that energy of body segments may change under such condition to compensate the force loss. For instance, under fatigue, it can be speculated that total and within segment energy of the body would increase, whilst the energy exchange between body segments would reduce. It may be an appealing strategy of a system to be more efficient under fatigue, losing less energy to the environment and using more energy within the system. The coordinative structure, rather than acting as a functional unit of many muscles, it is also a dissipated structure (Kelso *et al.*, 1980). It means that a structure promotes energy exchanges with the environment. Then, the speculative adjustment based on utilisation of energy may be a strategy of the coordinative structure would perform to compensate the force loss and maintain the required target height.

Despite of this speculative adjustment, there were adjustments in control parameters, such as increased joint flexion and jumping duration. These major adjustments were performed at I50 period and they did not change throughout the end of the session. One possible explanation is that the system is always searching for stability, irrespective of different conditions, at the same time that is flexible to allow adjustments in order to try to maintain the system stable. Then, muscle fatigue did not constrain the organism enough to induce changes on movement pattern, but was sufficient to induce adjustments to maintain the system stable. Another two aspects could also be argued, the effects of different work to rest ratio and cause of fatigue on movement pattern. Different work to rest ratio, manipulating the resting interval between jumps, produced a different number of countermovement jumps and exercise duration, thereby inducing central and peripheral fatigue. However, neither different intensity nor different causes of fatigue were able to produce changes in movement pattern of countermovement jump. It means that no matter to what extent neural input is arriving on exercised musculature, because the coordinative structure, and consequently the movement pattern, has not changed.

4.2 General conclusions

Changing one parameter of stretch-shortening cycle exercise fatigue protocol, such as the resting interval between countermovement jumps, is able to produce central and/or peripheral fatigue. It suggests that the cause of fatigue is task-dependent. Even with the neuromuscular system impaired, there was no change in the movement pattern of countermovement jump, irrespective of exercise duration or cause of fatigue. However, adjustments in control of countermovement jumps were seen in order to compensate the force loss and to maintain the targeted jump height. Therefore, the motor system remained stable under fatigue, performing few adjustments which suggest flexibility of the motor system to the conditions that the movement is performed.

4.3 Recommendations for future studies

It is recommended that future studies test the cause of fatigue in stretch-shortening cycle exercise performed continuously and intermittently. Disparities of results have been reported during isolated forms of muscle contraction, whilst stretch-shortening cycle exercise has not been investigated from this point of view. Moreover, numerous activities (e.g., team and racket sports, daily activities) are intermittent in nature and thus differ from laboratory testing (e.g., successive rebound jumps). Hereby, comparisons between intermittent and continuous activities remain an interesting topic that should be further investigated.

Future studies regarding the effects of musculoskeletal fatigue on movement pattern of multi-joint task are also recommended. However, the fatigue-induced exercise protocol must be interrupted at exhaustion instant, i.e., when participants would be unable to perform desired motor task. Determination of body and segment energies would be interesting to investigate the movement effectiveness as criterion for movement control in fatigued jumps. In addition, measurements and/or approaches of movement pattern closer to the motor control area, such as the uncontrolled manifold hypothesis, should be evaluated.

REFERENCES

Alkner, B. A., Tesch, P. A., & Berg, H. E. (2000). Quadriceps EMG/force relationship in knee extension and leg press. Med. Sci. Sports Exerc. 32, 459-463.

Avela, J., Finni, J., & Komi, P. V. (2006). Excitability of the soleus reflex arc during intensive stretch-shortening cycle exercise in two power-trained athlete groups. Eur. J. Appl. Physiol 97, 486-493.

Bampouras, T. M., Reeves, N. D., Baltzopoulos, V., & Maganaris, C. N. (2006). Muscle activation assessment: effects of method, stimulus number, and joint angle. Muscle Nerve 34, 740-746.

Bernstein, N. (1967). The coordination and regulation of movement. New York: Pergamon.

Bigland-Ritchie, B., Rice, C. L., Garland, S. J., & Walsh, M. L. (1995). Task-dependent factors in fatigue of human voluntary contractions. Adv. Exp. Med. Biol. 384, 361-380.

Bigland-Ritchie, B. & Woods, J. J. (1984). Changes in muscle contractile properties and neural control during human muscular fatigue. Muscle Nerve 7, 691-699.

Bilodeau, M. (2006). Central fatigue in continuous and intermittent contractions of triceps brachii. Muscle Nerve 34, 205-213.

Bobbert, M. F. & van Ingen Schenau, G. J. (1988). Coordination in vertical jumping. J Biomech. 21, 249-262.

Bobbert, M. F. & Van Soest, A. J. (1994). Effects of muscle strengthening on vertical jump height: a simulation study. Med. Sci. Sports Exerc. 26, 1012-1020.

Bogdanis, G. C., Nevill, M. E., Boobis, L. H., Lakomy, H. K., & Nevill, A. M. (1995). Recovery of power output and muscle metabolites following 30 s of maximal sprint cycling in man. J Physiol 482 (Pt 2), 467-480.

Bongiovanni, L. G. & Hagbarth, K. E. (1990). Tonic vibration reflexes elicited during fatigue from maximal voluntary contractions in man. J Physiol 423, 1-14.

Bonnard, M., Sirin, A. V., Oddsson, L., & Thorstensson, A. (1994). Different strategies to compensate for the effects of fatigue revealed by neuromuscular adaptation processes in humans. Neurosci. Lett. 166, 101-105.

Bosco, C. & Komi, P. V. (1979). Mechanical characteristics and fiber composition of human leg extensor muscles. Eur. J Appl Physiol Occup. Physiol 41, 275-284.

Clark, J. E., Philips, S. J., & Petersen, R. (1989). Developmental stability in jumping. Developmental Psychology 25, 929-935.

Cote, J. N., Mathieu, P. A., Levin, M. F., & Feldman, A. G. (2002). Movement reorganization to compensate for fatigue during sawing. Exp. Brain Res. 146, 394-398.

Day, S. J. & Hulliger, M. (2001). Experimental simulation of cat electromyogram: evidence for algebraic summation of motor-unit action-potential trains. J. Neurophysiol. 86, 2144-2158.

Di, G. C., Daniele, F., & Tipton, C. M. (2006). Angelo Mosso and muscular fatigue: 116 years after the first Congress of Physiologists: IUPS commemoration. Adv. Physiol Educ. 30, 51-57.

Dimitrova, N. A. & Dimitrov, G. V. (2003). Interpretation of EMG changes with fatigue: facts, pitfalls, and fallacies. J. Electromyogr. Kinesiol. 13, 13-36.

Drillis, R. J. & Contini, R. (1966). Technical Report, New York: New York University, 1166-03.

Duchateau, J. & Hainaut, K. (1985). Electrical and mechanical failures during sustained and intermittent contractions in humans. J. Appl. Physiol 58, 942-947.

Edwards, R. H. T. (1981). Human muscle function and fatigue, Human muscle fatigue: physiological mechanisms, pp. 1-18. London: Pitman Medical.

Enoka, R. M. & Stuart, D. G. (1992). Neurobiology of muscle fatigue. J Appl Physiol 72, 1631-1648.

Fitts, R. H. (1994). Cellular mechanisms of muscle fatigue. Physiol Rev. 74, 49-94.

Forestier, N. & Nougier, V. (1998). The effects of muscular fatigue on the coordination of a multijoint movement in human. Neurosci. Lett. 252, 187-190.

Gandevia, S. C. (2001). Spinal and supraspinal factors in human muscle fatigue. Physiol Rev. 81, 1725-1789.

Gibson, J. J. (1979). An ecological approach to visual perception. Boston: Houghton-Mifflin.

Hagbarth, K. E., Kunesch, E. J., Nordin, M., Schmidt, R., & Wallin, E. U. (1986). Gamma loop contributing to maximal voluntary contractions in man. J. Physiol 380, 575-591.

Hautier, C. A., Arsac, L. M., Deghdegh, K., Souquet, J., Belli, A., & Lacour, J. R. (2000). Influence of fatigue on EMG/force ratio and cocontraction in cycling. Med. Sci. Sports Exerc. 32, 839-843.

Horita, T., Komi, P. V., Nicol, C., & Kyrolainen, H. (1996). Stretch shortening cycle fatigue: interactions among joint stiffness, reflex, and muscle mechanical performance in the drop jump [corrected]. Eur. J. Appl. Physiol Occup. Physiol 73, 393-403.

Hortobagyi, T., Lambert, N. J., & Kroll, W. P. (1991). Voluntary and reflex responses to fatigue with stretch-shortening exercise. Can. J. Sport Sci. 16, 142-150.

Hudson, J. L. (1986). Coordination of segments in the vertical jump. Med. Sci. Sports Exerc. 18, 242-251.

Huffenus, A. F., Amarantini, D., & Forestier, N. (2006). Effects of distal and proximal arm muscles fatigue on multi-joint movement organization. Exp. Brain Res. 170, 438-447.

Kelso, J. A. S., Holt, K. G., Kugler, P. N., & Turvey, M. T. (1980). On the concept of coordenative structures as dissipative structures: II. Empirical lines of convergence, Tutorials in motor behaviour, edited by G. E. Stelmach & J. Requin, pp. 49-70. Amsterdam: North-Holand.

Komi, P. V. (2000). Stretch-shortening cycle: a powerful model to study normal and fatigued muscle. J. Biomech. 33, 1197-1206.

Kugler, P. N., Kelso, J. A. S., & Turvey, M. T. (1980). On the concept of coordenative structures as dissipative structures: I.Theoretical lines of convergence., Tutorials in motor behavior, edited by G. E. Stelmach & J. Requin, pp. 3-47. Amsterdan: North-Holland.

Kuitunen, S., Avela, J., Kyrolainen, H., & Komi, P. V. (2004). Voluntary activation and mechanical performance of human triceps surae muscle after exhaustive stretch-shortening cycle jumping exercise. Eur. J. Appl. Physiol 91, 538-544.

Kuitunen, S., Avela, J., Kyrolainen, H., Nicol, C., & Komi, P. V. (2002). Acute and prolonged reduction in joint stiffness in humans after exhausting stretch-shortening cycle exercise. Eur. J. Appl. Physiol 88, 107-116.

Lay, B. S., Sparrow, W. A., Hughes, K. M., & O'Dwyer, N. J. (2002). Practice effects on coordination and control, metabolic energy expenditure, and muscle activation. Hum. Mov Sci. 21, 807-830.

Lepers, R., Maffiuletti, N. A., Rochette, L., Brugniaux, J., & Millet, G. Y. (2002). Neuromuscular fatigue during a long-duration cycling exercise. J. Appl. Physiol 92, 1487-1493.

Linthorne, N. P. (2001). Analysis of standing vertical jumps using a force platform. American Journal of Physics 69, 1198-1204.

Merton, P. A. (1954). Voluntary strength and fatigue. J. Physiol 123, 553-564.

Metzger, J. M. & Moss, R. L. (1990). Effects of tension and stiffness due to reduced pH in mammalian fast- and slow-twitch skinned skeletal muscle fibres. J. Physiol 428, 737-750.

Millet, G. Y., Lepers, R., Maffiuletti, N. A., Babault, N., Martin, V., & Lattier, G. (2002). Alterations of neuromuscular function after an ultramarathon. J. Appl. Physiol 92, 486-492.

Millet, G. Y., Martin, V., Lattier, G., & Ballay, Y. (2003). Mechanisms contributing to knee extensor strength loss after prolonged running exercise. J Appl Physiol 94, 193-198.

Moritani, T., Oddson, L., & Thorstensson, A. (1990). Electromyographic evidence of selective fatigue during the eccentric phase of stretch/shortening cycles in man. Eur. J Appl Physiol Occup. Physiol 60, 425-429.

Newell, K. M. (1986). Constraints on the development of coordination, Motor development in children: aspects of coordination and control, edited by M. G. Wade & W. T. A. Whiting, pp. 341-360. Boston: Martinus Nijhoff.

Nicol, C., Avela, J., & Komi, P. V. (2006). The stretch-shortening cycle : a model to study naturally occurring neuromuscular fatigue. Sports Med. 36, 977-999.

Nicol, C., Komi, P. V., Horita, T., Kyrolainen, H., & Takala, T. E. (1996). Reduced stretch-reflex sensitivity after exhausting stretch-shortening cycle exercise. Eur. J. Appl. Physiol Occup. Physiol 72, 401-409.

Nicol, C., Komi, P. V., & Marconnet, P. (1991). Effects of marathon fatigue on running kinematics and economy. Scand J Med Sci Sports p. 195-204.

Nicol, C., Kuitunen, S., Kyrolainen, H., Avela, J., & Komi, P. V. (2003). Effects of long- and short-term fatiguing stretch-shortening cycle exercises on reflex EMG and force of the tendon-muscle complex. Eur. J. Appl. Physiol 90, 470-479.

Pereira, G. Efeito da duração do intervalo em saltos verticais: relações entre trabalho realizado e duração até a exaustão. Dissertação de Mestrado, Universidade Estadual Paulista, Rio Claro, 2004, p.134.

Pereira, G., Freitas Junior, P. B., & Kokubun, E. (2004). The critical power model fit to vertical jump. 9^{TH} Congress of European College of Sport Science p. 61.

Place, N., Maffiuletti, N. A., Martin, A., & Lepers, R. (2007). Assessment of the reliability of central and peripheral fatigue after sustained maximal voluntary contraction of the quadriceps muscle. Muscle Nerve.

Racinais, S., Girard, O., Micallef, J. P., & Perrey, S. (2007). Failed excitability of spinal motoneurons induced by prolonged running exercise. J Neurophysiol. 97, 596-603.

Reeves, N. D., Narici, M. V., & Maganaris, C. N. (2004). Effect of resistance training on skeletal muscle-specific force in elderly humans. J Appl Physiol 96, 885-892.

Regueme, S. C., Nicol, C., Barthelemy, J., & Grelot, L. (2005). Acute and delayed neuromuscular adjustments of the triceps surae muscle group to exhaustive stretch-shortening cycle fatigue. Eur. J. Appl. Physiol 93, 398-410.

Rodacki, A. L., Fowler, N. E., & Bennett, S. J. (2001). Multi-segment coordination: fatigue effects. Med. Sci. Sports Exerc. 33, 1157-1167.

Rodacki, A. L., Fowler, N. E., & Bennett, S. J. (2002). Vertical jump coordination: fatigue effects. Med. Sci. Sports Exerc. 34, 105-116.

SENIAM (1999). European recommendations for surface ElectroMyoGraphy edited by H. Hermens, B. Freriks, R. Merletti, D. Stegeman, J. Blok, G. Rau, C. Disselhorst-Klug, & G. Häg, The Netherlands: RRD.

Skurvydas, A., Dudoniene, V., Kalvenas, A., & Zuoza, A. (2002). Skeletal muscle fatigue in long-distance runners, sprinters and untrained men after repeated drop jumps performed at maximal intensity. Scand. J. Med. Sci. Sports 12, 34-39.

Skurvydas, A., Jascaninas, J., & Zachovajevas, P. (2000). Changes in height of jump, maximal voluntary contraction force and low-frequency fatigue after 100 intermittent or continuous jumps with maximal intensity. Acta Physiol Scand. 169, 55-62.

Sparto, P. J., Parnianpour, M., Reinsel, T. E., & Simon, S. (1997). The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting test. J Orthop. Sports Phys. Ther. 25, 3-12.

Strojnik, V. & Komi, P. V. (1998). Neuromuscular fatigue after maximal stretch-shortening cycle exercise. J Appl Physiol 84, 344-350.

Strojnik, V. & Komi, P. V. (2000). Fatigue after submaximal intensive stretch-shortening cycle exercise. Med. Sci. Sports Exerc. 32, 1314-1319.

Turvey, M. T. (1990). Coordination. Am. Psychol. 45, 938-953.

van Ingen Schenau, G. J., Van Soest, A. J., Gabreels, F. J. M., & Horstink, M. W. I. M. (1995). The control of multi-joint movements relies on detailed internal representations pp. 531-538.

Vanrenterghem, J., Bobbert, M. F., Casius, L. J., & De, C. D. (2006). Is energy expenditure taken into account in human sub-maximal jumping? - A simulation study. J Electromyogr. Kinesiol.

Vanrenterghem, J., Lees, A., Lenoir, M., Aerts, P., & De, C. D. (2004). Performing the vertical jump: movement adaptations for submaximal jumping. Hum. Mov Sci. 22, 713-727.

Ward, C. W., Spangenburg, E. E., Diss, L. M., & Williams, J. H. (1998). Effects of varied fatigue protocols on sarcoplasmic reticulum calcium uptake and release rates. Am. J Physiol 275, R99-R104.

Winter, D. A. (2005). Biomechanics and motor control of human movement, 3 ed. Waterloo: Wiley.

Yao, W., Fuglevand, R. J., & Enoka, R. M. (2000). Motor-unit synchronization increases EMG amplitude and decreases force steadiness of simulated contractions. J Neurophysiol. 83, 441-452.

RESUMO

O objetivo principal desta tese foi investigar a causa e a conseqüência da fadiga em *countermovement jumps* realizados em diferentes razões de esforço e pausa. Esses objetivos foram testados utilizando dois estudos experimentais. O primeiro estudo comparou as causas da fadiga em protocolos de fadiga de curta (=10 min) e longa duração (=20 min), manipulando as durações da pausa entre *countermovement jumps*. Os resultados indicaram que, após a realização do protocolo de fadiga de curta duração, houve redução da contração isométrica voluntária máxima para extensão do joelho acompanhada por fadiga central e periférica. Por outro lado, após a realização do protocolo de fadiga de longa duração, a redução da contração isométrica voluntária máxima para extensão do joelho foi associada apenas à fadiga periférica. No segundo estudo foi utilizado um protocolo de fadiga para o sistema neuromuscular similar ao primeiro estudo, tendo como objetivo investigar o padrão de movimento sob fadiga. O tempo e a seqüência da relação inter-segmental do movimento e da ativação muscular permaneceram inalteradas sob a condição de fadiga. Entretanto, alguns ajustes no controle de *countermovement jumps* realizados sob a condição de fadiga foram observados, provavelmente na tentativa de manter a altura do salto vertical. Entre alguns ajustes, é possível destacar um aumento na flexão das articulações e na duração do salto vertical. Esses resultados foram similares independentemente da razão de esforço e pausa utilizada para induzir fadiga em *countermovement jumps*. Considerando ambos os estudos experimentais, os resultados da presente tese sugerem que a causa da fadiga em *countermovement jumps* é tarefa-dependente. Isso significa que dependendo do protocolo de indução de fadiga utilizado, a causa da fadiga pode ser alterada de central para periférica, por exemplo. Mesmo o sistema neuromuscular tendo sua capacidade de realizar trabalho reduzido, não houve alteração no padrão de movimento de *countermovement jumps*, independente da duração do exercício ou da causa da fadiga. Porém, foram observados ajustes no controle de *countermovement jumps* na tentativa de compensar a redução da força muscular e manter a altura do salto vertical.

Palavras-chave: fadiga central, fadiga periférica, coordenação de movimento, controle de movimento, ciclo de alongamento-encurtamento.

APPENDIX A

```
% Sample Rate (Hz)
smprt_1=1000; % LabView
smprt_2=200; % Optotrak

% Defining some variables that must be fixed according to maximal jump height
bm=72.09; % body mass (kg)
bw = bm*9.81; % body weight (N)
f_sl = 0.132; % foot segment length (m)
s_sl = 0.4045; % shank segment length (m)
t_sl = 0.402; % thigh segment length (m)
h_sl = 0.52; % hat segment length (m)

% Loading the files
% Asking information about participant trial
sub=input('Enter with the name and trial number to be analyzed related to optotrak (i.e. e4_1) : ', 's');
trial=input('Enter with the name and trial number to be analyzed related to LabView (i.e. eb4s_1): ', 's');

% Preparing and loading the force plate file
data = [trial, '.txt'];
eval(['load ' data]);

% Defining the variables on columns related to LabView
eval(['fz_t= ' trial ' (: , 5) ; ']); % total vertical forces (Fz1, Fz2, Fz3 and Fz4)
eval(['fy_14= ' trial ' (: , 8) ; ']); % anterior force 1_4
eval(['fy_23= ' trial ' (: , 9) ; ']); % anterior force 2_3
eval(['emg_ga= ' trial ' (: , 10) ; ']); % emg of gastro muscle
eval(['emg_vl= ' trial ' (: , 11) ; ']); % emg of vastus lateralis muscle
eval(['emg_sm= ' trial ' (: , 12) ; ']); % emg of semitendinosus muscle
eval(['emg_vm= ' trial ' (: , 13) ; ']); % emg of vastus medialis muscle

% Sum of longitudinal forces on platform (fy)
fy_t = fy_14 + fy_23;

% Preparing and loading the optotrak file
data1 = [sub, '.txt'];
eval(['load ' data1]);

% Defining columns kinematic variables
eval(['vt_mt= ' sub ' (: , 2) ; ']); % metatarsal vertical
eval(['ap_mt= ' sub ' (: , 3) ; ']); % metatarsal antero-posterior
eval(['vt_ml= ' sub ' (: , 5) ; ']); % malleolus vertical oscillation
eval(['ap_ml= ' sub ' (: , 6) ; ']); % malleolus antero-posterior
eval(['vt_kn= ' sub ' (: , 8) ; ']); % knee vertical oscillation
eval(['ap_kn= ' sub ' (: , 9) ; ']); % knee antero-posterior oscillation
eval(['vt_hp= ' sub ' (: , 11) ; ']); % hip vertical oscillation
eval(['ap_hp= ' sub ' (: , 12) ; ']); % hip antero-posterior oscillation
```

```
eval(['vt_ac= ' sub ' (: , 14) ; ']); % acromion vertical oscillation
eval(['ap_ac= ' sub ' (: , 15) ; ']); % acromion antero-posterior

% Defining parameters to smooth the data
fq_cut=5;      % low pass cut frequency (EMG)
fq_cut_p=13;  % low pass cut frequency (Force Platform)
fq_cut_o=5;       % low pass cut frequency (Opto)
n = 2;              % smooth order
Wn = fq_cut/(1000/2);       % defining cut frequency for EMG
Wn1 = fq_cut_p/(1000/2);  % defining cut frequency for Force Plate
Wn2= fq_cut_o/(200/2);   % defining cut frequency for Opto
[b,a] = butter(n,Wn);    % Butterworth filter EMG
[d,c] = butter(n,Wn1);   % Butterworth filter Force Plate
[f,e] = butter(n,Wn2);   % Butterworth filter Optotrak

% Full-wave rectification of EMG data
emg_ga = abs(emg_ga);
emg_vl = abs(emg_vl);
emg_sm = abs(emg_sm);
emg_vm = abs(emg_vm);

% Smoothing all variables with different parameters (defined previously by residual
analysis) for kinematic and kinetic data. Linear envelope was applied to EMG data
(low-pass filter, second order, 5 Hz)

% Smooth the force plate and EMG data
fz=filtfilt(d,c,fz_t);
fy=filtfilt(d,c,fy_t);
emgga=filtfilt(b,a,emg_ga);
emgvl=filtfilt(b,a,emg_vl);
emgsm=filtfilt(b,a,emg_sm);
emgvm=filtfilt(b,a,emg_vm);

% Smooth the Optotrak data
vtmt=filtfilt(f,e,vt_mt);
apmt=filtfilt(f,e,ap_mt);
vtml=filtfilt(f,e,vt_ml);
apml=filtfilt(f,e,ap_ml);
vtkn=filtfilt(f,e,vt_kn);
apkn=filtfilt(f,e,ap_kn);
vthp=filtfilt(f,e,vt_hp);
aphp=filtfilt(f,e,ap_hp);
vtac=filtfilt(f,e,vt_ac);
apac=filtfilt(f,e,ap_ac);

% The horizontal oscillations of the all markers are multiplied by (-1) to negative and
positive values meant fore-after directions respectively
apmt = apmt.*(-1);
```

```
apml = apml.*(-1);
apkn = apkn.*(-1);
aphp = aphp.*(-1);
apac = apac.*(-1);

% Transferring all kinematic data to positive x and y square (mm). Finding the highest absolute value in vertical and horizontal kinematic data
max_vtmt=max(abs(vtmt));
max_apmt=max(abs(apmt));
max_vtml=max(abs(vtml));
max_apml=max(abs(apml));
max_vtkn=max(abs(vtkn));
max_apkn=max(abs(apkn));
max_vthp=max(abs(vthp));
max_aphp=max(abs(aphp));
max_vtac=max(abs(vtac));
max_apac=max(abs(apac));

% Creating a new matrix with maximal values, finding the max value on new matrix, and add 10
max_vt = [max_vtmt max_vtml max_vtkn max_vthp max_vtac];
max_vt = max(max_vt);
max_vt = max_vt+10;
max_ap = [max_apmt max_apml max_apkn max_aphp max_apac];
max_ap = max(max_ap);
max_ap = max_ap+10;

% Adding 10 upon max value of kinematic data
vtmt = vtmt+max_vt;
vtml = vtml+max_vt;
vtkn = vtkn+max_vt;
vthp = vthp+max_vt;
vtac = vtac+max_vt;
apmt = apmt+max_ap;
apml = apml+max_ap;
apkn = apkn+max_ap;
aphp = aphp+max_ap;
apac = apac+max_ap;

% Defining the Center of Pressure (m)
cop_y = apmt./1000;
cop_z = (vtmt(600,1))/1000;

% Calculating angles, velocities and accelerations of segments and joints. Velocities and accelerations are smoothed, as well. These calculations have used funlast1
f1=funlast1(vtmt, vtml, vtkn, vthp, vtac, apmt, apml, apkn, aphp, apac);

function result1=funlast1(vtmt, vtml, vtkn, vthp, vtac, apmt, apml, apkn, aphp, apac);
```

```
ed=length(vtmt);
% Estimating Segment Angles
% Foot Segment Angle
f_vt_ap = ((vtml-vtmt)./(apml-apmt)); % dividing Z/Y oscillations
f_rad = atan(f_vt_ap);                  % estimating foot angle in rad
f_deg = (abs(f_rad.*180/pi));     % transforming foot angle (rad) to absolute value in degree
% Shank Segment Angle
s_vt_ap = ((vtkn-vtml)./(apkn-apml));
for n = 1:ed; % Loop to estimate shank segment angle (rad)
  if apkn(n) > apml(n);
  s_rad(n) = atan(s_vt_ap(n));
    else if apkn(n) < apml(n);
  s_rad(n) = pi + (atan(s_vt_ap(n)));
    else if apkn(n) > apml(n);
  s_rad(n) = atan(s_vt_ap(n));
  else if apkn(n) < apml(n);
  s_rad(n) = pi + (atan(s_vt_ap(n)));
end
end
end
end
end
s_rad = transpose(s_rad); % Shank Segment Angle (rad)
s_deg = s_rad.*180/pi;    % Shank Segment Angle (degree)
% Thigh Segment Angle
t_vt_ap = ((vthp-vtkn)./(aphp-apkn));
for n = 1:ed; % Loop to estimate thigh segment angle (rad)
  if aphp(n) > apkn(n);
  t_rad(n) = atan(t_vt_ap(n));
    else if aphp(n) < apkn(n);
  t_rad(n) = pi + (atan(t_vt_ap(n)));
    else if aphp(n) > apkn(n);
  t_rad(n) = atan(t_vt_ap(n));
  else if aphp(n) < apkn(n);
  t_rad(n) = pi + (atan(t_vt_ap(n)));
end
end
end
end
end
t_rad = transpose(t_rad); % Thigh Segment Angle (rad)
t_deg = t_rad.*180/pi;    % Thigh Segment Angle (degree)
% Hat Segment Angle
h_vt_ap = ((vtac-vthp)./(apac-aphp));
for n = 1:ed; % Loop to estimate hat segment angle (rad)
  if apac(n) > aphp(n);
```

```
  h_rad(n) = atan(h_vt_ap(n));
    else if apac(n) < aphp(n);
  h_rad(n) = pi + (atan(h_vt_ap(n)));
    else if apac(n) > aphp(n);
  h_rad(n) = atan(h_vt_ap(n));
    else if apac(n) < aphp(n);
  h_rad(n) = pi + (atan(h_vt_ap(n)));
end
end
end
end
end
h_rad = transpose(h_rad); % Hat Segment Angle (rad)
h_deg = h_rad.*180/pi;    % Hat Segment Angle (degree)

% Estimating Joint Angles (degree and rad)
ankle_angle = f_deg + s_deg;          % degree
ankle_angle_r = ankle_angle./180*pi;   % rad
knee_angle = (180-t_deg) + (s_deg);    % degree
knee_angle_r = knee_angle./180*pi;     % rad
hip_angle = (180-t_deg) + (h_deg);     % degree
hip_angle_r = hip_angle./180*pi;       % rad

% Estimating joint angle displacement (deg) to calculate stiffness
% Knee angle displacement
knee_angle_disp = diff(knee_angle);
% Ankle angle displacement
ankle_angle_disp = diff(ankle_angle);
% Hip angle displacement
hip_angle_disp = diff(hip_angle);

% Defining parameters to smooth velocity and acceleration data
fq_cut_v = 7;  % low pass cut frequency for velocity and acceleration
n = 2;             % smooth order
Wn3 = fq_cut_v/(200/2);  % Defining cut frequency velocity and acceleration
[h,g] = butter(n,Wn3);        % Butterworth filter to velocity and acceleration

% Estimating Segment Angular Velocity and Acceleration (rad/s), followed by
smoothing.
deltat = (1/200)*2; % time interval between two frames used to estimate velocity and
acceleration
% Velocity
f_sav = (diff(f_rad))/deltat; % Foot segment angular velocity
s_sav = (diff(s_rad))/deltat; % Shank segment angular velocity
t_sav = (diff(t_rad))/deltat; % Thigh segment angular velocity
h_sav = (diff(h_rad))/deltat; % Hat segment angular velocity
f_savf=filtfilt(h,g,f_sav);
s_savf=filtfilt(h,g,s_sav);
```

```
t_savf=filtfilt(h,g,t_sav);
h_savf=filtfilt(h,g,h_sav);

% Acceleration
f_saa = (diff(f_savf))/deltat; % Foot segment angular acceleration
s_saa = (diff(s_savf))/deltat; % Shank segment angular acceleration
t_saa = (diff(t_savf))/deltat; % Thigh segment angular acceleration
h_saa = (diff(h_savf))/deltat; % Hat segment angular acceleration
f_saaf=filtfilt(h,g,f_saa);
s_saaf=filtfilt(h,g,s_saa);
t_saaf=filtfilt(h,g,t_saa);
h_saaf=filtfilt(h,g,h_saa);

% Estimating Joint Angular Velocity and Acceleration (rad/s), followed by smoothing.
% Velocity
ankle_jav = (diff(ankle_angle_r))/deltat; % Ankle joint angular velocity
knee_jav = (diff(knee_angle_r))/deltat; % Knee joint angular velocity
hip_jav = (diff(hip_angle_r))/deltat; % Hip joint angular velocity

% Knee Angular Velocity in deg/s to calculate the Time Transition between eccentric
and concentric phase
knee_jav_deg = (diff(knee_angle))/deltat;
ankle_javf=filtfilt(h,g,ankle_jav);
knee_javf=filtfilt(h,g,knee_jav);
knee_javf_deg=filtfilt(h,g,knee_jav_deg);
hip_javf=filtfilt(h,g,hip_jav);

% Acceleration
ankle_jaa = (diff(ankle_javf))/deltat; % Ankle joint angular acceleration
knee_jaa = (diff(knee_javf))/deltat; % Knee joint angular acceleration
hip_jaa = (diff(hip_javf))/deltat; % Knee joint angular acceleration
ankle_jaaf=filtfilt(h,g,ankle_jaa);
knee_jaaf=filtfilt(h,g,knee_jaa);
hip_jaaf=filtfilt(h,g,hip_jaa);

% Cutting variables length to have the same size
eda = length(hip_jaaf);
ankle_angle = ankle_angle(1:eda);
ankle_angle_r = ankle_angle_r(1:eda);
knee_angle = knee_angle(1:eda);
knee_angle_r = knee_angle_r(1:eda);
hip_angle = hip_angle(1:eda);
hip_angle_r = hip_angle_r(1:eda);
f_savf = f_savf(1:eda);
s_savf = s_savf(1:eda);
t_savf = t_savf(1:eda);
h_savf = h_savf(1:eda);
ankle_javf = ankle_javf(1:eda);
```

```
knee_javf = knee_javf(1:eda);
hip_javf = hip_javf(1:eda);
knee_javf_deg = knee_javf_deg(1:eda);
knee_angle_disp = knee_angle_disp(1:eda);
ankle_angle_disp = ankle_angle_disp(1:eda);
hip_angle_disp = hip_angle_disp(1:eda);

% Saving only interested variables
result1=[ankle_angle ankle_angle_r knee_angle knee_angle_r hip_angle hip_angle_r
f_saaf s_saaf t_saaf h_saaf f_savf s_savf t_savf h_savf ankle_jaaf knee_jaaf hip_jaaf
ankle_javf knee_javf hip_javf knee_javf_deg knee_angle_disp ankle_angle_disp
hip_angle_disp];

ankle_angle = f1(: , 1);
ankle_angle_r = f1(: , 2);
knee_angle = f1(: , 3);
knee_angle_r = f1(: , 4);
hip_angle = f1(: , 5);
hip_angle_r = f1(: , 6);
f_saaf = f1(: , 7);
s_saaf = f1(: , 8);
t_saaf = f1(: , 9);
h_saaf = f1(: , 10);
f_savf = f1(: , 11);
s_savf = f1(: , 12);
t_savf = f1(: , 13);
h_savf = f1(: , 14);
ankle_jaaf = f1(: , 15);
knee_jaaf = f1(: , 16);
hip_jaaf = f1(: , 17);
ankle_javf = f1(: , 18);
knee_javf = f1(: , 19);
hip_javf = f1(: , 20);
knee_javf_deg = f1(: , 21);
knee_angle_disp = f1(: , 22);
ankle_angle_disp = f1(: , 23);
hip_angle_disp = f1(: , 24);

% Calculating accelerations and velocities of segment center of mass on vertical and
fore-aft direction using funlast2
f2=funlast2(vtmt, vtml, vtkn, vthp, vtac, apmt, apml, apkn, aphp, apac);

function result2=funlast2(vtmt, vtml, vtkn, vthp, vtac, apmt, apml, apkn, aphp, apac);

% Defining parameters to smooth velocity and acceleration data
fq_cut_v = 7;   % low pass cut frequency for velocity and acceleration
n = 2;          % smooth order
Wn3 = fq_cut_v/(200/2);  % Defining cut frequency velocity and acceleration
```

```
[h,g] = butter(n,Wn3);        % Butterworth filter to velocity and acceleration

% Estimating the Segment Center of Mass (m) to Vertical (z) and Fore-Aft (y)
Directions. The values at the end of equations (i.e. 0.626) come from Winter (1990)
hat_comy = (apac./1000)+(((aphp-apac)./1000).*0.626);
hat_comz = (vtac./1000)-(((vtac-vthp)./1000).*0.626);
thigh_comy = (aphp./1000)+(((apkn-aphp)./1000).*0.433);
thigh_comz = (vthp./1000)-(((vthp-vtkn)./1000).*0.433);
shank_comy = (apkn./1000)+(((apml-apkn)./1000).*0.433);
shank_comz = (vtkn./1000)-(((vtkn-vtml)./1000).*0.433);
foot_comy = (apml./1000)+(((apmt-apml)./1000).*0.5);
foot_comz = (vtml./1000)-(((vtml-vtmt)./1000).*0.5);

% Estimating Center of Mass Velocity (m/s) and Acceleration (m/s2) to Vertical (z) and
Fore-Aft (y) Directions for all Body Segments, followed by smoothing
% Velocity
deltat = (1/200)*2; % time interval between two frames used to estimate velocity and
acceleration
hat_comy_v = (diff(hat_comy))/deltat; % Center of mass velocity of Hat on fore-aft
hat_comz_v = (diff(hat_comz))/deltat; % Center of mass velocity of Hat on vertical
thigh_comy_v = (diff(thigh_comy))/deltat; % Center of mass velocity of Thigh on fore-
aft
thigh_comz_v = (diff(thigh_comz))/deltat; % Center of mass velocity of Thigh on
vertical
shank_comy_v = (diff(shank_comy))/deltat; % Center mass velocity of Shank on fore-
aft
shank_comz_v = (diff(shank_comz))/deltat; % Center of mass velocity of Shank on
vertical
foot_comy_v = (diff(foot_comy))/deltat; % Center of mass velocity of Foot on fore-aft
foot_comz_v = (diff(foot_comz))/deltat; % Center of mass velocity of Foot on vertical
hat_comy_vf = filtfilt(h,g,hat_comy_v);
hat_comz_vf = filtfilt(h,g,hat_comz_v);
thigh_comy_vf = filtfilt(h,g,thigh_comy_v);
thigh_comz_vf = filtfilt(h,g,thigh_comz_v);
shank_comy_vf = filtfilt(h,g,shank_comy_v);
shank_comz_vf = filtfilt(h,g,shank_comz_v);
foot_comy_vf = filtfilt(h,g,foot_comy_v);
foot_comz_vf = filtfilt(h,g,foot_comz_v);

% Acceleration
hat_comy_a = (diff(hat_comy_vf))/deltat; % Center of mass acceleration of Hat on fore-
aft
hat_comz_a = (diff(hat_comz_vf))/deltat; % Center of mass acceleration of Hat on
vertical
thigh_comy_a = (diff(thigh_comy_vf))/deltat; % Center mass acceleration of Thigh
fore-aft
thigh_comz_a = (diff(thigh_comz_vf))/deltat; % Center of mass acceleration Thigh
vertical
```

```
shank_comy_a = (diff(shank_comy_vf))/deltat; % Center mass acceleration Shank fore-
aft
shank_comz_a = (diff(shank_comz_vf))/deltat; % Center mass acceleration Shank
vertical
foot_comy_a = (diff(foot_comy_vf))/deltat; % Center of mass acceleration Foot on
fore-aft
foot_comz_a = (diff(foot_comz_vf))/deltat; % Center of mass acceleration Foot on
vertical
hat_comy_af = filtfilt(h,g,hat_comy_a);
hat_comz_af = filtfilt(h,g,hat_comz_a);
thigh_comy_af = filtfilt(h,g,thigh_comy_a);
thigh_comz_af = filtfilt(h,g,thigh_comz_a);
shank_comy_af = filtfilt(h,g,shank_comy_a);
shank_comz_af = filtfilt(h,g,shank_comz_a);
foot_comy_af = filtfilt(h,g,foot_comy_a);
foot_comz_af = filtfilt(h,g,foot_comz_a);

% Cutting variables length to have the same size
eda = length(foot_comz_af);
hat_comy = hat_comy(1:eda);
hat_comz = hat_comz(1:eda);
thigh_comy = thigh_comy(1:eda);
thigh_comz = thigh_comz(1:eda);
shank_comy = shank_comy(1:eda);
shank_comz = shank_comz(1:eda);
foot_comy = foot_comy(1:eda);
foot_comz = foot_comz(1:eda);
hat_comy_vf = hat_comy_vf(1:eda);
hat_comz_vf = hat_comz_vf(1:eda);
thigh_comy_vf = thigh_comy_vf(1:eda);
thigh_comz_vf = thigh_comz_vf(1:eda);
shank_comy_vf = shank_comy_vf(1:eda);
shank_comz_vf = shank_comz_vf(1:eda);
foot_comy_vf = foot_comy_vf(1:eda);
foot_comz_vf = foot_comz_vf(1:eda);

% Saving only interested variables
result2=[hat_comy hat_comz thigh_comy thigh_comz shank_comy shank_comz
foot_comy foot_comz hat_comy_af hat_comz_af thigh_comy_af thigh_comz_af
shank_comy_af shank_comz_af foot_comy_af foot_comz_af hat_comy_vf
hat_comz_vf thigh_comy_vf thigh_comz_vf shank_comy_vf shank_comz_vf
foot_comy_vf foot_comz_vf];

hat_comy = f2(: , 1);
hat_comz = f2(: , 2);
thigh_comy = f2(: , 3);
thigh_comz = f2(: , 4);
shank_comy = f2(: , 5);
```

```
shank_comz = f2(: , 6);
foot_comy = f2(: , 7);
foot_comz = f2(: , 8);
hat_comy_af = f2(: , 9);
hat_comz_af = f2(: , 10);
thigh_comy_af = f2(: , 11);
thigh_comz_af = f2(: , 12);
shank_comy_af = f2(: , 13);
shank_comz_af = f2(: , 14);
foot_comy_af = f2(: , 15);
foot_comz_af = f2(: , 16);
hat_comy_vf = f2(: , 17);
hat_comz_vf = f2(: , 18);
thigh_comy_vf = f2(: , 19);
thigh_comz_vf = f2(: , 20);
shank_comy_vf = f2(: , 21);
shank_comz_vf = f2(: , 22);
foot_comy_vf = f2(: , 23);
foot_comz_vf = f2(: , 24);

% Plotting Fz to fix the jump length to be analyzed using Ginput function
x=(1:length(fz));
figure(1)
plot(fz);
hold on;
title('Instants of Jumping')
xlabel('Time (ms)')
ylabel('Fz (N)')
[in_x,in_y]=ginput(1);
[fn_x,fn_y]=ginput(1);

% Calculating the variables related to force traces using funlast3
f3=funlast3(in_x, fn_x, fz, bm);

function result3=funlast3(in_x, fn_x, fz, bm);
bw = bm*9.81; % body weight (N)
x=(1:length(fz));

bw_5=bw-(bw.*0.05); % defining the value 5% below body weight
p_a_f=find(fz(round(in_x:fn_x))<bw_5); % finding values lower than pc_5 on fz curve
p_a=in_x+(p_a_f(1,1)-1); % defining the "a" instant (starting eccentric phase)
p_c_f=find(fz(round(p_a:fn_x))>bw); % finding values greater than pc after "a" instant
p_c=p_a+(p_c_f(1,1)-1); % defining the "c" instant (fz force is greater than body weight)
p_e_f=find(fz(round(p_c:fn_x))<bw); % finding values lower than pc after "c" instant
p_e=p_c+(p_e_f(1,1)-1); % defining the "e" instant (fz force is lower than body weight)
p_f_f=find(fz(round(in_x:fn_x))<0); % finding values lower than zero on fz curve
p_f=in_x+(p_f_f(1:1)-1); % defining the "f" instant (take off instant)
```

```
% Calculating unweighting phase of CMJ
quad1=trapz((x(round(p_a):round(p_c))./1000),linspace(fz(round(p_a)),fz(round(p_c)),
length(round(p_a):round(p_c)))); % total area between"a"and"c"instants
area_b=trapz((x(round(p_a):round(p_c))./1000),fz(round(p_a):round(p_c))); %
calculating area below fz curve between "a"  and "c" instants
uw = quad1-area_b;  % unweighting value (N)

% Calculating the instant which the CMJ change from eccentric to concentric and its
respectives time durations
j=round(p_c)+1;
for k = j:round(p_f);
area_ex1(k)=(trapz((x(round(p_c):k)/1000),fz(round(p_c):k))); % the area between "c"
and "f" instants
area_quad(k)=(fz(j-1)*(x(k-round(p_c))/1000)); % calculating the square area below pc
area_ex(k)=area_ex1(k)-area_quad(k); % the positive work between "c" end "f" instants
  end
t_e=find((area_ex(round(p_c):round(p_f))) > uw)'; % finding the area above pc and
greater than uw value
t_ex = round(p_c) + t_e(2,1); % defining the transition instant from eccentric to
concentric
t_exc = ((t_ex-1)-p_a)/1000; % defining the eccentric time duration (s)
t_con = (p_f-t_ex)/1000; % defining the concentric time duration (s)
t_exccon = t_exc + t_con; % defining total jumping duration (s)

% Calculating the concentric phase and force peak
quad2=fz(round(p_e))*((p_e-p_c)./1000);  % body weight area between instants "c" and
"e"
quadn=trapz((x(round(p_e:p_f))./1000), fz(round(p_e:p_f))); % area that quad2 didn't
do
quadt=quad2+quadn; % the total area below body weight between instants "c" and "f"
area_ac=trapz((x(round(p_c:p_f))./1000),fz(round(p_c:p_f))); %total area between "c"
and "f"
area_p=area_ac-quadt; % the total area above body weight between instants "c" and "e"
imp_con = area_p-uw; % the concentric impulse (N)
pkf = (max(fz(round(in_x:p_e)))).(bw); % the force peak/body weight

 % Calculating the jump height by moment-impulse method
area_1=(((p_f-p_a)/1000).*bw); % the body weight area between instants "a" and "f"
area=trapz((x(round(p_a:p_f))/1000),fz(round(p_a:p_f))); % the area below fz curve
between instants "a" and "f"
area2=area-area_1; % Determining the area to calculate Voff
v_off = ((area2)./(bw./9.81)); % Calculating Voff
h_jump = ((v_off^2)./(2*9.81)); % Jump height (m)

% Saving only interested variables
result3=[p_a p_c p_f uw t_ex t_exc t_con imp_con pkf h_jump t_exccon p_e];

p_a = f3(: , 1);
```

```
p_c = f3(: , 2);
p_f = f3(: , 3);
uw = f3(: , 4);
t_ex = f3(: , 5);
t_exc = f3(: , 6);
t_con = f3(: , 7);
imp_con = f3(: , 8);
pkf = f3(: , 9);
h_jump = f3(: , 10)
t_exccon = f3(: , 11);
p_e = f3(: , 12);

plot(x(round(p_a)), fz(round(p_a)), 'mo');
hold on;
plot(x(round(p_c)), fz(round(p_c)), 'yo');
hold on;
plot(x(round(t_ex)), fz(round(t_ex)), 'g*');
hold on;
plot(x(round(p_f)), fz(round(p_f)), 'co');

% Reducing force plate data from 1000 to 200 Hz
red = length(fz)/5;
for m = 1:red;
  fz_red(m) = fz(((m-1)*5)+1);
end
fz_red = transpose(fz_red);

for m = 1:red;
  fy_red(m) = fy(((m-1)*5)+1);
end
fy_red = transpose(fy_red);
fz_red = fz_red(1:length(hat_comy_af));
fy_red = fy_red(1:length(hat_comy_af));
vtmt = vtmt(1:length(hat_comy_af));
vtml = vtml(1:length(hat_comy_af));
vtkn = vtkn(1:length(hat_comy_af));
vthp = vthp(1:length(hat_comy_af));
vtac = vtac(1:length(hat_comy_af));
apmt = apmt(1:length(hat_comy_af));
apml = apml(1:length(hat_comy_af));
apkn = apkn(1:length(hat_comy_af));
aphp = aphp(1:length(hat_comy_af));
apac = apac(1:length(hat_comy_af));
cop_y = cop_y(1:length(hat_comy_af));

% Defining instants over force curve after data reducing
p_ar = (p_a/5)+1;
p_cr = (p_c/5)+1;
```

```
p_er = (p_e/5)+1;
p_fr = (p_f/5)+1;
t_exr = (t_ex/5)+1;

xred=1:length(hat_comy_af);
xred = xred-1;
xred = transpose(xred);

figure(2)
plot(xred,fz_red)
hold on
title('Instants of Jumping With Fz Reduced')
plot(xred(round(p_ar)), fz_red(round(p_ar)), 'mo');
xlabel('Time (ms)')
ylabel('Fz (N)')
hold on;
plot(xred(round(p_cr)), fz_red(round(p_cr)), 'yo');
hold on;
plot(xred(round(t_exr)), fz_red(round(t_exr)), 'g*');
hold on;
plot(xred(round(p_fr)), fz_red(round(p_fr)), 'co');
hold on;

figure(3)
vtmt_v = diff(vtmt)/((1/200)*2);
xv=1:length(vtmt_v);
plotyy(xv,vtmt_v,xv,fz_red(1:end-1))
title('Synchronization')
xlabel('Time (ms)')
legend('Joint Velocity (rad/s)', 'Vertical Force (N)', 2)

% Calculating the EMG peak (mV) and time difference (ms) from take off for eccentric and concentric phase
% Eccentric phase
[pk_ga_ecc,x_pk_ga_ecc] = max(emgga(round(p_a:t_ex-1)));
td_pkoff_ga_ecc = (p_a + x_pk_ga_ecc) - (p_f);
[pk_sm_ecc,x_pk_sm_ecc] = max(emgsm(round(p_a:t_ex-1)));
td_pkoff_sm_ecc = (p_a + x_pk_sm_ecc) - (p_f);
[pk_vl_ecc,x_pk_vl_ecc] = max(emgvl(round(p_a:t_ex-1)));
td_pkoff_vl_ecc = (p_a + x_pk_vl_ecc) - (p_f);
[pk_vm_ecc,x_pk_vm_ecc] = max(emgvm(round(p_a:t_ex-1)));
td_pkoff_vm_ecc = (p_a + x_pk_vm_ecc) - (p_f);
% Concentric phase
[pk_ga_con,x_pk_ga_con] = max(emgga(round(t_ex:p_f)));
td_pkoff_ga_con = ((t_ex) + x_pk_ga_con) - (p_f);
[pk_sm_con,x_pk_sm_con] = max(emgsm(round(t_ex:p_f)));
td_pkoff_sm_con = ((t_ex) + x_pk_sm_con) - (p_f);
[pk_vl_con,x_pk_vl_con] = max(emgvl(round(t_ex:p_f)));
```

```
td_pkoff_vl_con = ((t_ex) + x_pk_vl_con) - (p_f);
[pk_vm_con,x_pk_vm_con] = max(emgvm(round(t_ex:p_f)));
td_pkoff_vm_con = ((t_ex) + x_pk_vm_con) - (p_f);

% Calculating the iEMG (mV.s) and iEMG average (mV) for eccentric and concentric phase
% Eccentric phase
iemgga_ec = (trapz((x(round(p_a:t_ex-1))./smprt_1), emgga(round(p_a:t_ex-1))));
iemgga_ec_av = iemgga_ec/t_exc;
iemgsm_ec = (trapz((x(round(p_a:t_ex-1))./smprt_1), emgsm(round(p_a:t_ex-1))));
iemgsm_ec_av = iemgsm_ec/t_exc;
iemgvl_ec = (trapz((x(round(p_a:t_ex-1))./smprt_1), emgvl(round(p_a:t_ex-1))));
iemgvl_ec_av = iemgvl_ec/t_exc;
iemgvm_ec = (trapz((x(round(p_a:t_ex-1))./smprt_1), emgvm(round(p_a:t_ex-1))));
iemgvm_ec_av = iemgvm_ec/t_exc;
% Concentric phase
iemgga_co = (trapz((x(round(t_ex:p_f))./smprt_1), emgga(round(t_ex:p_f))));
iemgga_co_av = iemgga_co/t_con;
iemgsm_co = (trapz((x(round(t_ex:p_f))./smprt_1), emgsm(round(t_ex:p_f))));
iemgsm_co_av = iemgsm_co/t_con;
iemgvl_co = (trapz((x(round(t_ex:p_f))./smprt_1), emgvl(round(t_ex:p_f))));
iemgvl_co_av = iemgvl_co/t_con;
iemgvm_co = (trapz((x(round(t_ex:p_f))./smprt_1), emgvm(round(t_ex:p_f))));
iemgvm_co_av = iemgvm_co/t_con;

% Determining when the initiation of muscle activation occurs before take off. This instant corresponds to the first frame after the muscle activation had increased 20% of the activation peak value in concentric phase
ga20 = (pk_ga_con*20)/100;
emgga20 = find(emgga(round(p_a:p_f))>ga20);
in_act_ga = ((p_a+(emgga20(1,1)))-p_f); % gastrocnemius
sm20 = (pk_sm_con*20)/100;
emgsm20 = find(emgsm(round(p_a:p_f))>sm20);
in_act_sm = ((p_a+(emgsm20(1,1)))-p_f); % semitendinosus initiation
vl20 = (pk_vl_con*20)/100;
emgvl20 = find(emgvl(round(p_a:p_f))>vl20);
in_act_vl = ((p_a+(emgvl20(1,1)))-p_f); % vastus lateralis
vm20 = (pk_vm_con*20)/100;
emgvm20 = find(emgvm(round(p_a:p_f))>vm20);
in_act_vm = ((p_a+(emgvm20(1,1)))-p_f); % vastus medialis

% Relative Mass of Segment (kg). The values multiplied by body mass coming from Winters book (1990)
hat_smr = bm*0.678;
thigh_smr = bm*0.1;
shank_smr = bm*0.0465;
foot_smr = bm*0.0145;
```

```
% Moment of Inertia (Io) on Center of Mass (kg.m^2). The values multiplied by segment
length come from Winter (1990)
hat_Io = (hat_smr)*((h_sl*0.496)^2);
thigh_Io = (thigh_smr)*((t_sl*0.323)^2);
shank_Io = (shank_smr)*((s_sl*0.302)^2);
foot_Io = (foot_smr)*((f_sl*0.475)^2);

% Estimating the Joint Net Moments (Nm)
% Ankle
ray=((foot_comy_af(round(p_ar-tdifc:p_fr-tdifc))).*(foot_smr))-
(fy_red(round(p_ar:p_fr)));  % Ankle fore-aft reaction force
raz=((foot_comz_af(round(p_ar-tdifc:p_fr-tdifc))).*(foot_smr))+(foot_smr*9.81)-
(fz_red(round(p_ar:p_fr))); % Ankle vertical reaction force
ma = (f_saaf(round(p_ar-tdifc:p_fr-tdifc)).*foot_Io)-((ray).*((vtml(round(p_ar-
tdifc:p_fr-tdifc))./1000)-(foot_comz(round(p_ar-tdifc:p_fr-tdifc)))))-
((raz).*((apml(round(p_ar-tdifc:p_fr-tdifc))./1000)-(foot_comy(round(p_ar-tdifc:p_fr-
tdifc)))))-((fy_red(round(p_ar:p_fr))).*((cop_z)-(foot_comz(round(p_ar-tdifc:p_fr-
tdifc)))))-((fz_red(round(p_ar:p_fr))).*((cop_y(round(p_ar-tdifc:p_fr-tdifc)))-
(foot_comy(round(p_ar-tdifc:p_fr-tdifc)))));
ma_c = ma.*(-1); % it was multiplied by -1 to correct its direction

% Knee
rky = ((shank_comy_af(round(p_ar-tdifc:p_fr-tdifc))).*(shank_smr))+(ray); % Knee
fore-aft reaction force
rkz=((shank_comz_af(round(p_ar-tdifc:p_fr-
tdifc))).*(shank_smr))+(raz)+(shank_smr*9.81);  % Knee vertical reaction force
mk=((s_saaf(round(p_ar-tdifc:p_fr-tdifc))).*(shank_Io))-(-ma)-((-
ray).*((shank_comz(round(p_ar-tdifc:p_fr-tdifc)))-(vtml(round(p_ar-tdifc:p_fr-
tdifc))./1000)))-((-raz).*((apml(round(p_ar-tdifc:p_fr-tdifc))./1000)-
(shank_comy(round(p_ar-tdifc:p_fr-tdifc)))))-((rky).*((shank_comz(round(p_ar-
tdifc:p_fr-tdifc)))-(vtkn(round(p_ar-tdifc:p_fr-tdifc))./1000)))-
((rkz).*((apkn(round(p_ar-tdifc:p_fr-tdifc))./1000)-(shank_comy(round(p_ar-tdifc:p_fr-
tdifc))))); % Net Moment about Knee Joint

% Hip
rhy = ((thigh_comy_af(round(p_ar-tdifc:p_fr-tdifc))).*(thigh_smr))+(rky); % Hip fore-
aft reaction force
rhz= (thigh_comz_af(round(p_ar-tdifc:p_fr-
tdifc))).*(thigh_smr))+(rkz)+(thigh_smr*9.81); % Hip vertical reaction force
mh = ((t_saaf(round(p_ar-tdifc:p_fr-tdifc))).*(thigh_Io))-(-mk)-((-
rky).*((vtkn(round(p_ar-tdifc:p_fr-tdifc))./1000)-(thigh_comz(round(p_ar-tdifc:p_fr-
tdifc)))))-((-rkz).*((apkn(round(p_ar-tdifc:p_fr-tdifc))./1000)-(thigh_comy(round(p_ar-
tdifc:p_fr-tdifc)))))-((rhy).*((vthp(round(p_ar-tdifc:p_fr-tdifc))./1000)-
(thigh_comz(round(p_ar-tdifc:p_fr-tdifc)))))-((rhz).*((aphp(round(p_ar-tdifc:p_fr-
tdifc))./1000)-(thigh_comy(round(p_ar-tdifc:p_fr-tdifc)))));
mh_c = mh.*(-1); % it was multiplied by -1 to correct its direction

figure(4)
```

```
plot([(apmt(round(100-tdifc))), (apml(round(100-tdifc))), (apkn(round(100-tdifc))),
(aphp(round(100-tdifc))),(apac(round(100-tdifc)))],[(vtmt(round(100-
tdifc))),(vtml(round(100-tdifc))),(vtkn(round(100-tdifc))),(vthp(round(100-tdifc))),
(vtac(round(100-tdifc)))], 'y')
axis equal
hold on;
title('With Synchronisation')
plot([(apmt(round(p_ar+40-tdifc))), (apml(round(p_ar+40-tdifc))),
(apkn(round(p_ar+40-tdifc))),(aphp(round(p_ar+40-tdifc))),(apac(round(p_ar+40-
tdifc)))], [(vtmt(round(p_ar+40-tdifc))),(vtml(round(p_ar+40-tdifc))),
(vtkn(round(p_ar+40-tdifc))), (vthp(round(p_ar+40-tdifc))),(vtac(round(p_ar+40-
tdifc)))])
axis equal
hold on;
plot([(apmt(round(p_ar+80-tdifc))), (apml(round(p_ar+80-tdifc))),
(apkn(round(p_ar+80-tdifc))),(aphp(round(p_ar+80-tdifc))),(apac(round(p_ar+80-
tdifc)))], [(vtmt(round(p_ar+80-tdifc))),(vtml(round(p_ar+80-tdifc))),
(vtkn(round(p_ar+80-tdifc))), (vthp(round(p_ar+80-tdifc))),(vtac(round(p_ar+80-
tdifc)))])
axis equal
hold on;
plot([(apmt(round(t_exr-tdifc))), (apml(round(t_exr-tdifc))), (apkn(round(t_exr-tdifc))),
(aphp(round(t_exr-tdifc))),(apac(round(t_exr-tdifc)))],(vtmt(round(t_exr-tdifc))),
(vtml(round(t_exr-tdifc))),(vtkn(round(t_exr-tdifc))),(vthp(round(t_exr-tdifc))),
(vtac(round(t_exr-tdifc)))], 'r')
axis equal
hold on;
plot([(apmt(round(t_exr+20-tdifc))), (apml(round(t_exr+20-tdifc))),
(apkn(round(t_exr+20-tdifc))),(aphp(round(t_exr+20-tdifc))),(apac(round(t_exr+20-
tdifc)))], (vtmt(round(t_exr+20-tdifc))),(vtml(round(t_exr+20-tdifc))),
(vtkn(round(t_exr+20-tdifc))), (vthp(round(t_exr+20-tdifc))),(vtac(round(t_exr+20-
tdifc)))], 'g')
axis equal
hold on
plot([(apmt(round(t_exr+40-tdifc))), (apml(round(t_exr+40-tdifc))),
(apkn(round(t_exr+40-tdifc))),(aphp(round(t_exr+40-tdifc))),(apac(round(t_exr+40-
tdifc)))], [(vtmt(round(t_exr+40-tdifc))),(vtml(round(t_exr+40-
tdifc))),(vtkn(round(t_exr+40-tdifc))), (vthp(round(t_exr+40-
tdifc))),(vtac(round(t_exr+40-tdifc)))], 'g')
axis equal
hold on
plot([(apmt(round(p_fr-tdifc))), (apml(round(p_fr-tdifc))), (apkn(round(p_fr-tdifc))),
(aphp(round(p_fr-tdifc))),(apac(round(p_fr-tdifc)))],[(vtmt(round(p_fr-tdifc))),
(vtml(round(p_fr-tdifc))),(vtkn(round(p_fr-tdifc))),(vthp(round(p_fr-tdifc))),
(vtac(round(p_fr-tdifc)))], 'mo--')
axis equal

figure(6)
```

```
subplot(4,1,1)
plot(fz_red(round(p_ar:p_fr)), 'r')
title('Vertical Force (N) and Joint Net Moment (N.m)')
ylabel('Vertical Force')
subplot(4,1,2)
plot(ma_c)
ylabel('Ankle')
subplot(4,1,3)
plot((mk), 'k')
ylabel('Knee')
subplot(4,1,4)
plot((mh_c), 'm')
ylabel('Hip')

% Peak of Net Moment (Nm) and time difference (ms) from take off for eccentric and concentric phase
xm = (1:length(ma))';
% Ankle
[pk_a_con,x_pk_a_m_con] = max(ma_c);
td_pkoff_a_m_con = x_pk_a_m_con - xm(end);
pk_a_m_con = pk_a_con - ma_c(1,1);
[pk_a_m_ecc,x_pk_a_m_ecc] = min(ma_c(1:x_pk_a_m_con)) ;
td_pkoff_a_m_ecc = x_pk_a_m_ecc - xm(end);
% Knee
[pk_k_con,x_pk_k_m_con] = max(mk);
td_pkoff_k_m_con = x_pk_k_m_con - xm(end);
pk_k_m_con = pk_k_con - mk(1,1);
[pk_k_m_ecc,x_pk_k_m_ecc] = min(mk(1:x_pk_k_m_con));
td_pkoff_k_m_ecc = x_pk_k_m_ecc - xm(end);
% Hip
[pk_h_con,x_pk_h_m_con] = max(mh_c);
td_pkoff_h_m_con = x_pk_h_m_con - xm(end);
pk_h_m_con = pk_h_con - mh_c(1,1);
[pk_h_m_ecc,x_pk_h_m_ecc] = min(mh_c(1:x_pk_h_m_con));
td_pkoff_h_m_ecc = x_pk_h_m_ecc - xm(end);

% Estimating the Power Peak around the Joints (Watts) for eccentric and concentric phases and the time difference to take off (ms)
ankle_power = (ma_c).*(ankle_javf(round(p_ar-tdifc:p_fr-tdifc)));
knee_power = (mk).*(knee_javf(round(p_ar-tdifc:p_fr-tdifc)));
hip_power = (mh_c).*(hip_javf(round(p_ar-tdifc:p_fr-tdifc)));
% Ankle
[pkp_ankle_con,x_pkp_ankle_con] = max(ankle_power);
tdif_pkoff_ankle_con = x_pkp_ankle_con - xm(end);
[pkp_ankle_ecc,x_pkp_ankle_ecc] = min(ankle_power(1:x_pkp_ankle_con));
tdif_pkoff_ankle_ecc = x_pkp_ankle_ecc - xm(end);
% Knee
[pkp_knee_con,x_pkp_knee_con] = max(knee_power);
```

```
tdif_pkoff_knee_con = x_pkp_knee_con - xm(end);
[pkp_knee_ecc,x_pkp_knee_ecc] = min(knee_power(1:x_pkp_knee_con));
tdif_pkoff_knee_ecc = x_pkp_knee_ecc - xm(end);
% Hip
[pkp_hip_con,x_pkp_hip_con] = max(hip_power);
tdif_pkoff_hip_con = x_pkp_hip_con - xm(end);
[pkp_hip_ecc,x_pkp_hip_ecc] = min(hip_power(1:x_pkp_hip_con));
tdif_pkoff_hip_ecc = x_pkp_hip_ecc - xm(end);

figure(7)
subplot(4,1,1)
plot(fz_red(round(p_ar:p_fr)), 'r')
title('Vertical Force (N) and Joint Powers (W)')
ylabel('Vertical Force')
subplot(4,1,2)
plot(ankle_power)
ylabel('Ankle')
subplot(4,1,3)
plot(knee_power, 'k')
ylabel('Knee')
subplot(4,1,4)
plot(hip_power, 'm')
ylabel('Hip')

% Estimating the Mechanical Work Done of Muscles (Joules) for eccentric and
concentric phase
tex_sync = (t_ex-p_a)/5; % Time duration of eccentric and concentric phases for data
reduced to 200Hz
tco_sync = (t_con*1000)/5;
ankle_work_ecc = trapz((xm(round(1:tex_sync-1))./200),
(ankle_power(round(1:tex_sync-1))));
ankle_work_con=trapz((xm(round(tex_sync):xm(end))./200),
(ankle_power(round(tex_sync):xm(end))));
knee_work_ecc = trapz((xm(round(1:tex_sync-1))./200),
(knee_power(round(1:tex_sync-1))));
knee_work_con=trapz((xm(round(tex_sync):xm(end))./200),
(knee_power(round(tex_sync):xm(end))));
hip_work_ecc = trapz((xm(round(1:tex_sync-1))./200), (hip_power(round(1:tex_sync-
1))));
hip_work_con=trapz((xm(round(tex_sync):xm(end))./200),
(hip_power(round(tex_sync):xm(end))));

% Total Work (eccentric + concentric) and its percentage of maximal
tot_ankle_work = (abs(ankle_work_ecc))+(abs(ankle_work_con));
tot_knee_work = (abs(knee_work_ecc))+(abs(knee_work_con));
tot_hip_work = (abs(hip_work_ecc))+(abs(hip_work_con));
total_work = tot_ankle_work + tot_knee_work + tot_hip_work;
perc_ankle_work = (tot_ankle_work*100)/total_work;
```

```
perc_knee_work = (tot_knee_work*100)/total_work;
perc_hip_work = (tot_hip_work*100)/total_work;
% Total work (eccentric)
tot_work_ecc = (abs(ankle_work_ecc))+(abs(knee_work_ecc))+(abs(hip_work_ecc));
% Positive Work (concentric) and its percentage of maximal
pos_work = ankle_work_con + knee_work_con + hip_work_con
perc_posw_ankle = (ankle_work_con*100)/pos_work;
perc_posw_knee = (knee_work_con*100)/pos_work;
perc_posw_hip = (hip_work_con*100)/pos_work;

% Calculating peak joint angular velocity (rad/s) and time difference to take off (ms)
[pk_ankle_jav,x_pk_ankle_jav] = max(ankle_javf(round(p_ar-tdifc:p_fr-tdifc)));
tdif_ankle_pkjav = (x_pk_ankle_jav) - xm(end);
[pk_knee_jav,x_pk_knee_jav] = max(knee_javf(round(p_ar-tdifc:p_fr-tdifc)));
tdif_knee_pkjav = (x_pk_knee_jav) - xm(end);
[pk_hip_jav,x_pk_hip_jav] = max(hip_javf(round(p_ar-tdifc:p_fr-tdifc)));
tdif_hip_pkjav = (x_pk_hip_jav) - xm(end);

figure(8)
 plot(xm,ankle_javf(round(p_ar-tdifc:p_fr-tdifc)),xm,knee_javf(round(p_ar-tdifc:p_fr-tdifc)),xm,hip_javf(round(p_ar-tdifc:p_fr-tdifc)), 'r')
title('Joint Angular Velocity');
xlabel('Time (s)')
ylabel('Joint Angular Velocity (rad/s)')
legend('Ankle', 'Knee', 'Hip', 2);

% Initiation of Extension (iex) for all joints and time difference (tdif) between iex and take off (ms)
[y_ankle_angle_min,x_ankle_angle_min] = min(ankle_angle(round(p_ar-tdifc:p_fr-tdifc)));
tdif_iex_toff_ankle = (x_ankle_angle_min+1) - xm(end);
[y_knee_angle_min,x_knee_angle_min] = min(knee_angle(round(p_ar-tdifc:p_fr-tdifc)));
tdif_iex_toff_knee = (x_knee_angle_min+1) - xm(end);
[y_hip_angle_min,x_hip_angle_min] = min(hip_angle(round(p_ar-tdifc:p_fr-tdifc)));
tdif_iex_toff_hip = (x_hip_angle_min+1) - xm(end);

% The relative timing and sequential relationship between adjacent segments. In other words, % calculating a proximal-to-distal order by difference between iex (ms)
delay_hip_knee = (x_hip_angle_min+1) - (x_knee_angle_min+1);
delay_hip_ankle = (x_hip_angle_min+1) - (x_ankle_angle_min+1);
delay_knee_ankle = (x_knee_angle_min+1) - (x_ankle_angle_min+1);

% Calculating Joint Reversal Time (ms). The reversion was calculated through the time interval between -30deg/s and +30deg/s from knee initiation of extension
knee_veltr = (knee_javf_deg(round(p_ar-tdifc:p_fr-tdifc)));
iex_knee_vel = (knee_veltr(x_knee_angle_min));
knee_vel_mais = (abs(iex_knee_vel))+30;
```

```
knee_vel_menos = (abs(iex_knee_vel))-30;
mais = find(knee_veltr > knee_vel_mais);
mais_30 = (x_knee_angle_min - (mais(1,1)-1));
[y_kjavf_deg,x_kjavf_deg] = min(knee_veltr);
menos = find(knee_veltr < knee_vel_menos);
menos_30 = (menos(end)+1) - x_knee_angle_min;
ttr = (abs(mais_30)) + (abs(menos_30)); % Time Transition (ms)

% Calculating Joint Stiffness (N.m/degree) during whole eccentric phase
% Ankle
delta_ma = (ma_c(xm(round(tex_sync-1))))-(ma_c(xm(1,1)));
delta_aangle_disp=(ankle_angle_disp(round(t_exr-1-tdifc)))-
(ankle_angle_disp(round(p_ar-tdifc)));
ankle_stiff = delta_ma/delta_aangle_disp;
% Knee
delta_mk = (mk(xm(round(tex_sync-1))))-(mk(xm(1,1)));
delta_kangle_disp= (knee_angle_disp(round(t_exr-1-tdifc)))-
(knee_angle_disp(round(p_ar-tdifc)));
knee_stiff = delta_mk/delta_kangle_disp;
% Hip
delta_mh = (mh_c(xm(round(tex_sync-1))))-(mh_c(xm(1,1)));
delta_hangle_disp = (hip_angle_disp(round(t_exr-1-tdifc)))-
(hip_angle_disp(round(p_ar-tdifc)));
hip_stiff = delta_mh/delta_hangle_disp;

% Calculating joint stiffness from the last 15 degrees of the negative phase to the
deepest excursion of the knee joint
% Ankle
ankle_angle_deepest_x = (p_ar-tdifc)+(x_ankle_angle_min);
ankle_angle_deepest_mais = y_ankle_angle_min+15;
menos_ankle_inst=find(ankle_angle(round(p_ar-
tdifc:ankle_angle_deepest_x))<ankle_angle_deepest_mais);
rev_ankle = ((p_ar-tdifc)+(menos_ankle_inst))-1;
rev_ankle_disp = diff(ankle_angle(round(rev_ankle)));
delta_ankle_disp = rev_ankle_disp(end,1) - rev_ankle_disp(1,1);
delt_ma_c=ma_c(round(menos_ankle_inst(end,1))-1)-
ma_c(round(menos_ankle_inst(1,1)-1));
ankle_stiff_15 = delt_ma_c/delta_ankle_disp;

% Knee
knee_angle_deepest_x = (p_ar-tdifc)+(x_knee_angle_min);
knee_angle_deepest_mais = y_knee_angle_min+15;
menos_knee_inst=find(knee_angle(round(p_ar-
tdifc:knee_angle_deepest_x))<knee_angle_deepest_mais);
rev_knee = ((p_ar-tdifc)+(menos_knee_inst))-1;
rev_knee_disp = diff(knee_angle(round(rev_knee)));
delta_knee_disp = rev_knee_disp(end,1) - rev_knee_disp(1,1);
```

```
delt_mk = mk(round(menos_knee_inst(end,1))-1) - mk(round(menos_knee_inst(1,1)-
1));
knee_stiff_15 = delt_mk/delta_knee_disp;

% Hip
hip_angle_deepest_x = (p_ar-tdifc)+(x_hip_angle_min);
hip_angle_deepest_mais = y_hip_angle_min+15;
menos_hip_inst=find(hip_angle(round(p_ar-
tdifc:hip_angle_deepest_x))<hip_angle_deepest_mais);
rev_hip = ((p_ar-tdifc)+(menos_hip_inst))-1;
rev_hip_disp = diff(hip_angle(round(rev_hip)));
delta_hip_disp = rev_hip_disp(end,1) - rev_hip_disp(1,1);
delt_mh_c = mh_c(round(menos_hip_inst(end,1))-1) -
mh_c(round(menos_hip_inst(1,1)-1));
hip_stiff_15 = delt_mh_c/delta_hip_disp;

% Saving Variables
variables =[uw; t_exc; t_con; t_exccon; imp_con; pkf; h_jump; iemgga_ec;
iemgga_ec_av; iemgsm_ec; iemgsm_ec_av; iemgvl_ec; iemgvl_ec_av; iemgvm_ec;
iemgvm_ec_av; iemgga_co; iemgga_co_av; iemgsm_co; iemgsm_co_av; iemgvl_co;
iemgvl_co_av; iemgvm_co; iemgvm_co_av; pkp_ankle_ecc; tdif_pkoff_ankle_ecc;
pkp_ankle_con; tdif_pkoff_ankle_con; pkp_knee_ecc; tdif_pkoff_knee_ecc;
pkp_knee_con; tdif_pkoff_knee_con; pkp_hip_ecc; tdif_pkoff_hip_ecc; pkp_hip_con;
tdif_pkoff_hip_con; ankle_work_ecc; ankle_work_con; knee_work_ecc;
knee_work_con; hip_work_ecc;      hip_work_con; tot_ankle_work; tot_knee_work;
tot_hip_work; total_work; perc_ankle_work; perc_knee_work; perc_hip_work;
pos_work; perc_posw_ankle; perc_posw_knee; perc_posw_hip; pk_ankle_jav;
tdif_ankle_pkjav;pk_knee_jav; tdif_knee_pkjav; pk_hip_jav; tdif_hip_pkjav;
delay_hip_knee; delay_hip_ankle; delay_knee_ankle; tdif_iex_toff_ankle;
tdif_iex_toff_knee; tdif_iex_toff_hip;     y_ankle_angle_min; y_knee_angle_min;
y_hip_angle_min; ankle_stiff; knee_stiff; hip_stiff; ankle_stiff_15; knee_stiff_15;
hip_stiff_15; ttr; pk_a_m_ecc; td_pkoff_a_m_ecc; pk_a_m_con; td_pkoff_a_m_con;
pk_k_m_ecc; td_pkoff_k_m_ecc; pk_k_m_con; td_pkoff_k_m_con; pk_h_m_ecc;
td_pkoff_h_m_ecc; pk_h_m_con; td_pkoff_h_m_con; pk_ga_ecc; td_pkoff_ga_ecc;
pk_ga_con;  td_pkoff_ga_con; pk_sm_ecc; td_pkoff_sm_ecc; pk_sm_con;
td_pkoff_sm_con; pk_vl_ecc; td_pkoff_vl_ecc; pk_vl_con; td_pkoff_vl_con;
pk_vm_ecc; td_pkoff_vm_ecc; pk_vm_con; td_pkoff_vm_con; in_act_ga; in_act_sm;
in_act_vl; in_act_vm; tot_work_ecc]';

% Saving file with LabView trial name
nome_arq_ascii = [trial, '.res'] ;

% Saving variables with ASCII format
eval(['save -ascii ', nome_arq_ascii , ' variables ']) ;
```