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AN ANALYSIS OF THE BIOMECHANICS OF LANDING OF TWO GROUPS OF ATHLETES

A Thesis in

Kinesiology

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ABSTRACT

Athletes, even at an elite level, will perform a skill in many different ways. Even though different methods may all be successful in respect to the sport, certain methods may predispose certain athletes to greater rates of injury. Females, in particular, experience an alarming rate of anterior cruciate ligament (ACL) injuries and at rates much higher than males in similar sports. The purpose of this study was to examine the biomechanics of landing from a drop for two groups of female athletes and to compare between the two groups their segmental inertial properties, landing kinematics and kinetics. Motion analysis, ground reaction forces, body segment inertial parameters and resultant joint moments of 8 varsity level female gymnasts and 8 varsity level swimmers performing ten two-footed landings onto a force plate from a nominal drop height of 0.35 meters were analyzed. For almost all body segments, the center of mass location, moment of inertia, and the length for each thigh, shank, and foot were not statistically significantly different between the two groups when normalized for body size. On landing the swimmers exhibited a statistically significant greater range of whole body center of mass motion; this was accompanied by greater ranges of motion at the ankle, knee and hip joints for this group. Almost all of the metrics for ground reaction forces were statistically significantly different between the groups, with the gymnasts having the greater values. Overall the gymnasts exhibited “stiffer” landings compared with the swimmers, in effect the gymnasts made less of an effort to dissipate the forces of the landing task than did the swimmers. The present study did have some limitations, primarily the use of motion analysis in one plane of motion. Future studies would

benefit from a full three-dimensional analysis and the inclusion of males to provide another point of comparison.

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CHAPTER 1

INTRODUCTION

1.1 General Introduction

Observations of athletes performing a similar activity, even with elite athletes, reveal there are a variety of ways in which the activity is executed. Whether these different movement patterns cause more effective outcomes, predispose subjects to injury, or maximizes the individuals abilities is hard to determine. The elucidation of the role of different movement patterns is the realm of biomechanics. Biomechanical analysis is time consuming so many activities of interest have not received the attention they perhaps warrant. The variety of movement patterns seen between athletes may indicate some different underlying mechanism at play.

One major athletic injury concern is the prevalence of non-contact anterior cruciate ligament (ACL) injuries, particularly for women (Arendt & Dick, 1995). Females have statistically suffered a higher percentage of these injuries than males. The prevalence of women having these injuries has been described as an epidemic and is a major concern for women participating in athletic activities. Jumping and agility programs, to train the neuromuscular system, have been proposed as a method for reducing these injuries in females (e.g., Herman et al., 2012), but studies are required so that good technique can be discerned from less favorable techniques (movement patterns).

1.2 Purpose of Study

The purpose of this study was to examine the biomechanics of landing from a drop for two groups of female athletes.

The specific aims of this study were to determine, for two groups of female subjects, gymnasts and swimmers:

1) Segmental inertial properties.
2) Center of mass motion during landings.
3) Ground reaction forces during the landings.
4) Lower limb joint kinematics during the landings.
5) Lower limb joint kinetics during the landings.
6) Statistically compare the inertial properties, landing kinematics and kinetics between the two groups.

1.3 Study Overview

To examine the difference in landing performance the two groups of subjects were asked to perform landings from a nominal drop height of 0.35 m. The analysis has combined force plate and motion analysis data to provide variables which describe these landings.

1.4 Thesis Structure

This thesis starts with a review of literature, Chapter 2; focusing on the kinematics and kinetics of landing from a drop. Chapter 3 introduces the methods used for the data collection, including specifications of the equipment used, the subjects of the study, and methods that were used for motion analysis, ground reaction force data, resultant joint moments, and to determine body segment inertial parameters. Chapter 4 reviews the data collected and details the statistical analysis with the use of descriptive tables and graphs filled with summarized data. The thesis concludes with Chapter 5, which discusses the findings of the thesis and the implications they have on both the literature and future studies.

CHAPTER 2

REVIEW OF THE LITERATURE

This chapter presents the relevant literature examining the kinematics and kinetics of landing from a drop. Sections one and two will cover traditional kinematic and kinetic analysis of a drop, respectively. Muscle activation sequencing patterns will be covered in section three, including a discussion of gender differences, or lack thereof. The chapter concludes with a summary.

2.1. Kinematics of Landing from a Drop

Two common assumptions are made when utilizing typical kinematic and kinetic analysis. The first of those presumes that body segments are rigid, that they do not bend, and that the tissue surrounding the underlying skeleton and therefore the segment center of mass position and moment of inertia are fixed. This assumption dictates that movement of the markers indicates movement of the underlying skeleton. This, however, does not account for skin or soft tissue movement, which could wrongly be viewed as skeletal movement. At impact, such as landing from a drop, soft-tissue movement shifts the position of the center of mass of a segment, and therefore whole body center of mass. This motion, however, is by definition ignored when inverse dynamics are carried out using a rigid body model. Rigid body models are created of objects after positional data is captured using image-based digitizing of body mounted marker locations. Velocity and acceleration are obtained by differentiating this positional data. Resultant joint moments, and other kinetic variables, are then computed utilizing this data. The error made from body segment inertial parameters (BSIP) estimation and positional data propagates with each subsequent calculation of kinematics and kinetics (Challis et al., 1996). The second common assumption is that all subjects drop from the same height since they leave the same height platform. Few studies have confirmed impact velocity and therefore drop height from impulse data. Comparisons of landing dynamics between groups or individuals are flawed if subjects drop from different drop heights by maneuvering their center of mass during leaving the jumping platform. Most kinematic analyses fail to attempt to identify these subtle differences.

2.1.1. Body and Joint Displacements

Several studies have examined the effects of varying drop height and surface stiffness on landing strategy. McNitt-Gray et al. (1993) attempted to determine if female gymnasts would vary their self-selected landing strategies when landing on mats of the same thickness, but varied composition, and from varied heights. Nine female gymnasts dropped from three different heights (0.69, 1.25 and 1.82 m) onto two different composition mats (soft and stiff). The results of the study showed that self-selected landing strategies of female gymnasts did differ under varying drop heights and mat compositions. As drop height increased, the degree of joint flexion, rate of joint flexion, impact peak magnitude, and landing phase duration all tended to increase. The stiffer mat also tended to elicit larger degrees of knee flexion. The differences in joint flexion and rate of joint flexion were smaller between different mat compositions as apposed to the different drop heights. This shows that the gymnasts in this study vary their landing strategies differently when adjusting for differences in height versus differences in mat composition. Several key assumptions, including bilateral symmetry, use of a rigid body model, skin-mounted marker movement, and nominal drop height reflecting actual drop height, were not considered or controlled for in any way.

McNitt-Gray (1991) used kinematic and ground reaction force data to analyze the effects of impact velocity and landing experience on “landing strategy”. Six male collegiate gymnasts and six male recreational athletes each performed four trials of dropping from three different height platforms (0.32, 0.72, and 1.28 m), associated with three landing velocities (2.5, 3.8, and 5.0 m/s respectively). Landings were barefooted with both feet theoretically landing simultaneously on a force plate. Two trials selected by each subject as a representation of their “preferred landing strategy” were used for analysis. Fifteen anatomical markers were used and digitized from the time of initial foot contact to the time when the whole body center of gravity (WBCG) reached a minimum vertical position. Results of the study showed mean landing phase duration increased (0.17, 0.26, 0.29 s) with increasing drop height for the recreational athletes, and decreased slightly (0.30, 0.29, 0.27 s) for the gymnasts. WBCG position at contact was not significantly different between the groups regardless of drop height. Both groups displayed lower WBCG positions with increased drop height. Angular position of the ankle, knee, and hip joint at contact were not significantly different for each group, suggesting subjects initiated contact in similar positions regardless of impact velocity. The gymnasts had slightly greater knee and ankle extensions at contact versus the recreational athletes. The recreational athletes also utilized greater hip flexion when landing from the higher heights and flexed the hip joint less from the low height and more from the higher height than the gymnasts.

McNitt-Gray et al. (1994) examined the influence of contact surface stiffness on the landing strategies of gymnasts. Ten female and four male intercollegiate gymnasts were used in the study. Eight of the ten female gymnasts were also used in the McNitt-Gray et al. (1993) study. Each subject performed four trials by stepping off a 0.69 m platform onto one of two mats (soft or stiff) covering a force plate or onto the force plate itself. Landing phase was defined from the time of initial contact to the time the subject brought the vertical velocity of the body center of gravity to zero. Results of the study indicated gymnasts modulated total body stiffness by scaling both hip and knee flexion in response to various landing surfaces. A greater degree of knee flexion was observed during landings onto stiffer surfaces. Females produced a greater maximum knee flexion angle and a slightly greater hip flexion angle at landing on average versus the males. Most of the females and all of the males had a greater degree of hip than knee flexion strategy. Confirmation of drop height was not attempted nor considered in any way.

Kovacs et al. (1999) compared heel-toe with forefoot landings from a drop. Subjects dropped from a 0.4 m platform placed one meter from a force plate. During heel-toe landings, vertical impact force peaked 20 to 25 ms after initial contact and landing duration (time of initial contact to time of subsequent jump takeoff) was 1.2 times longer than toe landings. Flexion phase duration (time from initial contact to maximum joint flexion position) was greatest for the knee compared with the ankle and hip regardless of landing style and was significantly greater for the heel-toe than forefoot landing conditions. Flexion phase duration was not significantly different at the hip during either landing condition, indicating subjects used similar hip flexion strategies regardless of landing style. Intuitively one would expect and kinematic analysis confirmed a greater degree of dorsi-flexion during heel-toe landings, and more plantar-flexion during forefoot landings. Regardless of landing condition, the rate of maximum hip joint flexion was consistently greater than maximum knee joint flexion.

Across all the mentioned studies, a few patterns and similarities are noticed. First is that an increase in drop height or surface stiffness led to an increase in landing phase duration. These increases also require a greater joint range of motion to control body descent. This also led to a longer recovery time. The increased joint range of motion is joint specific for each subject (e.g. some chose more hip flexion versus more knee or ankle flexion).

2.1.2. Joint Angular Velocities

Mean peak joint angular velocities for low, medium, and high angular velocities are presented in Table 2.1 for McNitt-Gray (1991). Mean maximum knee angular velocities of 618.8, 744.9, and 825.1 degrees/s for nine female gymnasts corresponding to drop heights of 0.69, 1.25 and 1.82 m respectively onto a stiff matt were reported in McNitt-Gray et al. (1993). In addition, peak hip angular velocities (412.0, 551.8, and 599.4 degrees/s) and landing phase duration (0.183, 0.218, and 0.230 s) increased with increasing drop height from 0.69 to 1.82 m. Differences in these values under changing matt stiffness were not significant. Again, several assumptions, and therefore possible sources of error, were not addressed, including rigid body model, bilateral symmetry, skin-mounted marker movement, and uniform drop height.

Table 2.1 - Mean (standard deviation) peak joint angular velocities for male gymnasts and recreational athletes. Velocities represent flexion at the joint (McNitt-Gray, 1991).

Group	Drop Height (m)	Ankle Angular Velocity (degrees/s)	Knee Angular Velocity (degrees/s)	Hip Angular Velocity (degrees/s)
Gymnasts	0.32	1059 (146)	646 (38)	424 (26)
	0.72	1244 (99)	771 (63)	576 (78)
	1.28	1403 (69)	942 (61)	697 (64)
Recreational Athletes	0.32	959 (167)	525 (59)	306 (90)
	0.72	1160 (126)	718 (78)	552 (100)
	1.28	1351 (151)	912 (104)	707 (81)

McNitt-Gray et al. (1994) reported greater knee and hip flexion in gymnasts, along with shorter times to peak knee angular velocities as contact surface stiffness increased. Peak hip angular velocities were not significantly effected, however. These results suggested that a greater amount of joint motion was utilized in an effort to dissipate additional impact forces during stiff or no mat conditions. Also, female gymnasts tended to have shorter landing phase duration than males across all mat conditions. Finally, results showed that gymnasts used a knee flexion strategy more on stiff surfaces compared with padded surfaces.

2.1.3. Common Assumptions

Studies examining the kinematics of landing from a drop or drop jump typically make several assumptions, including a rigid body model, and ignoring skin-mounted marker movement and soft tissue movement, despite the fact these assumptions affect all subsequent calculations (McNitt-Gray et al., 1993; McNitt-Gray et al., 1994; Kovacs et al., 1999). The only study to address the issue of drop height variability was Minetti et al. (1998). Using vertical ground reaction force impulse calculation, it suggested the raised platform step-off technique showed less variability than the vertical hang technique. Experimental results showed low variability, however, the difference between platform height and actual height dropped was substantial as summarized in Table 2.2.

Table 2.2 - Comparison of platform height and computed drop height from impulse data for sedentary and athletic populations as reported in Minetti et al., (1998), actual drop heights are means (standard deviations).

Group	Platform Height (m)	Actual Height Dropped (m)
Sedentary Males	0.40	0.332 (0.038)
Sedentary Males	0.71	0.554 (0.037)
Athletic Males	0.75	0.539 (0.092)
Athletic Females	0.75	0.461 (0.052)
Sedentary Males	1.10	0.806 (0.069)

Females were only included in the subject groups of three studies (McNitt-Gray et al., 1993, 1994, and Minetti et al., 1998). The only specific gender comparison made by McNitt-Gray et al. (1994) was that females demonstrated longer landing phase duration. Trends in their paper also indicated no significant differences in landing strategy chosen. Minetti et al. (1998) combined male and female muscle activity data as input for their simple mass-spring model, therefore gender differences were not considered when they predicted safe drop heights for an athletic population.

2.2. Kinetics of Landing from a Drop

People can choose among numerous mechanisms, some active while others passive, when dissipating impact forces experienced during landing from a drop. These forces must be dissipated or absorbed in order to protect the skeleton, along with the underlying organs and soft tissue, from a potentially damaging magnitude of force. The following sub-sections will review soft tissue motion and shock absorption, body position and landing strategy, typical ground reaction forces, resultant joint moments, and common assumptions in kinetic analysis.

2.2.1. Soft Tissue Motion and Shock Absorption

Passive shock absorption is achieved primarily through bone and soft tissue deformation. Several passive mechanisms contribute to shock absorption during landing from a drop onto the heels, including heel pad compression, bone bending, ligament and muscle stretching, cartilage deformation, and soft tissue motion. Zatsiorsky et al. (1987) reported that passive structures contribute substantially (from 0 to 75%) to force dissipation during “stiff” landings. This is perhaps evident during the oscillating motion of the leg soft tissue that occurs at heel strike during running. Zatsiorsky et al. (1987) determined the resultant joint moments from the kinematic data of a single subject performing either “very stiff” or “very soft” landings from 0.2 and 0.5 m. Passive energy dissipation mechanisms (e.g. heel pad deformation, soft tissue motion), regardless of the evidence about their contributions to shock absorption, are generally ignored in most kinetic analysis.

Andrews and Dowling (2000) used a fourth order mass-spring-damper mechanical model to estimate axial tibial accelerations resulting from impulsive heel impacts. A 50 G capacity uni-axial accelerometer (mass 18.2 g) was pre-loaded to 45 N and mounted on the skin over the medial tibial condyle. Fourteen subjects each performed three barefoot, knee fully extended landings onto the right heel pad. Landings were completed from a 0.5 m platform. The model consisted of two masses. The first was a single foot/leg segment mass (6.1% body mass). The second mass represented the rest of the body above the knee. Spring-damper systems linked each mass, and a spring represented the heel mass. Ten thousand possible stiffness and dampening coefficients were manipulated until an optimal match between estimated and measured tibial accelerations was found. Peak tibial accelerations from the model were within 0.5% of actual values, while predicted vertical ground reaction forces were lower and slightly more variable than measured. Relative differences in predicted tibial accelerations and rates of accelerations increased progressively when subject and group mean coefficients were utilized compared with individual subject and trial data.

2.2.2. Body Position and Landing Strategy

Mizrahi et al. (1982) examined the influences of body position, range of flexion of the lower extremity joints, softness of the ground, and the use of a ground roll immediately on impact during landing. Five subjects (two female and three male) performed vertical drops from a 0.5 and 1.0 m hang onto co-laterally mounted force plates. Each subject landed with one foot on each force plate. Subjects performed three drops from the lower height with 2 landing variations (onto the balls of the feet and flat footed). From the 1.0 m height, three subjects each performed four drops onto the balls of the feet. Two of these drops were followed by a lateral ground-roll. Two subjects also performed “soft” landings onto a 5 cm thick foam mat. Gender was not reported in the last two sets of trials. Flat footed landings produced a greatest first peak vertical force magnitudes. Landings onto the mat and those followed by a ground roll produced lower vertical force peaks. These results emphasize the role of muscle activation and joint movements in reducing peak forces during landings. Simultaneous use of all joints is probably the basis for reducing peak forces at impact.

Devita et al. (1992) had eight female basketball and volleyball players each perform ten landings from a vertical drop of 0.59 m. They compared vertical ground reaction forces, joint positions, joint moments, and muscle powers in the lower extremity during two types of landings, soft (greater than 90 degrees knee flexion) and stiff (less than 90 degrees knee flexion). During each trial, the subject landed onto both feet simultaneously with the right foot contacting a force plate. Vertical ground reaction force and film data were sampled at 1000 and 100 Hz, respectively. Although not directed to do so, subjects favored a forefoot landing style. Soft (117 degrees of knee flexion) landings resulted in a larger range of motion at each joint when compared to the stiff (77 degrees of knee flexion) landings. Stiff landings produced greater hip and knee resultant joint moments during the descent phase. This produced a more erect posture and flexed knee position at impact. Stiff landings had greater vertical ground reaction forces. Maximum vertical ground reaction force values occurred 12 and 50 ms after initial foot contact. Recovery times (time from initial contact to the instant when the vertical ground reaction force was an absolute or relative minimum) were 130 and 88 ms for soft and stiff landings respectively. In the soft landing condition, the hip and knee joints were identified as absorbing more energy. The ankle joint reportedly absorbed more energy during the stiff landing. The relative joint contributions to joint work for the hip, knee and ankle were 25, 37, and 37% for soft landings compared with 20, 31, and 50% for stiff landings. It was concluded that the ankle plantorflexors provided the majority of energy absorption in both landing conditions.

McNitt-Gray (1993) used male gymnasts and recreational athletes to compare lower extremity kinetics of barefoot drop landings from three heights (0.32, 0.72, and 1.28 m). Two-dimensional kinetic and kinematic data were collected and joint motions and ground reaction force data were used to calculate net joint forces, resultant joint moments, joint powers, and estimated work done at the knee and hip. As velocity of impact increased so did the net peak resultant joint moments and work done. Impact velocities of the whole body center of gravity (WBCG) were 2.5, 3.8, and 5.0 m/s for the low, medium, and high drop heights, respectively. The gymnasts exhibited larger peak hip (7.3 to 23.1 Nm/kg) and ankle (4.2 to 7.6 Nm/kg) resultant joint moments that increased with increasing height compared with the recreational athletes (7.7 to 17.5 and 4.2 to 6.2 Nm/kg for hip and ankle, respectively). Peak resultant joint moments for the knee joint did not appear significantly different between the groups. It was suggested that gymnasts appear to dissipate the impact forces by using larger ankle and hip moments as the impact velocity increased without increasing landing phase duration. Compared with the gymnasts, the recreational athletes exhibited greater hip resultant joint moments with longer landing phase duration as impact velocity increased. These results uphold the common assumption that gymnasts would be better able to control their descent from a drop across impact velocities when compared to recreational athletes.

2.2.3. Typical Ground Reaction Forces

Valiant et al. (1985) had male basketball players perform rebound jumps, wearing shoes, landing both feet simultaneously with their right foot striking a force plate. They reported mean first vertical ground reaction force peaks for forefoot landers of 1.3 BW and mean second peaks of 4.1 BW. Flatfoot landers averaged 6 BW on their single peak vertical ground reaction forces. Steele et al.

(1987) reported barefoot landings resulted in peak vertical ground reaction forces 1.1 times greater than shod landings. Adrian et al. (1983) had female collegiate volleyball players land from a moving block. They observed mean peak vertical ground reaction forces corresponding to 3.7 BW. Panzer et al. (1988) had gymnast perform double back somersaults landing simultaneously on both feet, with only the right foot contacting a force plate. They reported ground reaction forces of 12.3 and 15.1 BW. Kovacs et al. (1999) determined that mean first vertical ground reaction force peaks were 3.4 times larger for heel-toe than the toe only landing condition.

Studies of gymnasts and recreational athletes dropping from 0.32, 0.72 and 1.28 m platforms reported that gymnasts experienced greater impact peaks (3.9, 6.3, and 11.0 BW) compared with recreational athletes (4.2, 6.0, and 9.1 BW). The gymnasts reached these peaks an average of 6.3 ms earlier in the landing phase (McNitt-Gray, 1989). Mizrhai et al. (1982) observed greater vertical ground reaction forces (3.68 to 6.15 BW) for landings from 0.5 m onto flat feet as compared with the first vertical ground reaction force peaks from landing onto the toes (1.67 to 2.22 BW). Gymnasts performing barefooted vertical drops from two heights with various feedback instructions to land as soft as possible were able to reduce their vertical ground reaction force peaks by 50 and 63% for the respective drop heights of 0.25 and 0.30 m, by exhibiting greater forward trunk, hip, and knee flexion (Tant et al., 1989).

Similar findings were seen in McNitt-Gray (1993). Peak vertical ground reaction forces for a low (0.69) and medium (1.25 m) drop heights were similar between gymnasts (3.9 and 6.3 BW) and recreational athletes (4.2 and 6.4 BW). However, peaks for the higher drop height (1.82 m) were greater for gymnasts (11 BW) than the recreational athletes (9.1 BW). These values were significantly different between drop heights, but not between groups. There were several limitations to this study, however. Only the gymnasts preferred landings were analyzed, the assumption of a rigid body model was used for the calculation of resultant joint moments, and drop heights were not confirmed via impulse calculation.

2.2.4. Resultant Joint Moments

The consistency of mean resultant joint moments reported in the literature appears to be dependant on the method of data analysis (e.g. pooling or averaging data across subjects or groups before performing inverse dynamics), simplifying assumptions, and methods of calculation (Dufek et al., 1991). McNitt-Gray (1993) reported larger peak hip (7.3-23.1 Nm/kg) and ankle (4.2-7.6Nm/kg) resultant joint moments for gymnasts compared with recreational athletes (7.7-17.5 and 4.2-6.2 Nm/kg). There was no significant differences noted in the work done at the hip. Resultant joint moment peaks for the knee (1.5-7.6 Nm/kg) were not significantly different between the groups.

Kovacs et al. (1999) also examined resultant joint moments at the ankle, knee, and hip during the descent landing phase of a drop jump. The sum of those moments was reported to be similar between heel (389.4 ± 89.5 Nm) and toe (393.0 ± 11.3 Nm) landing conditions, with heel landing style resulting in 1.3 and 1.1 times greater mean resultant joint moment at the hip and knee.

2.2.5. Common Assumptions in Kinetic Analysis

In the study of biomechanics, the body is often assumed to be rigid. The validity of this assumption is questionable, however, in situations where high accelerations occur. Karlsson et al. (1994) reported soft tissue motion, evident from differences between skin and bone mounted marker movement on the thigh, during even slow movements. This movement is even more pronounced as a result of high frequency impacts (Lafortune et al., 1992). The assumption that a body is rigid when it is not increases the potential for error in both the kinematics and BSIP, and thus the kinetics.

Skin mounted markers tend to move independently of the joints over which they are placed. This results in a misrepresentation of the movement of the underlying skeleton and will hinder the accurate reconstruction of two and three-dimensional kinematic coordinates. This difference in motion between bone and skin is normally referred to as skin movement artifact. Again, this difference in actual versus recorded body movement can produce significant error in movement anaylsis, especially in activities involving high accelerations and impacts (Cappozzo et al., 1996).

Several studies of late have attempted to quantify the difference in motion between the skin and the skeleton and the error this difference can produce. The hope is that by determining what difference there is, it can be removed, thus improving anaylsis. Significant skin movement artifacts have been discovered for such activities as knee flexion and extension (Lafortune et al., 1992), walking and cycling (Cappozzo et al., 1996; Holden et al., 1994), and voluntary knee rotations (Karlsson et al., 1994).

Reinschmidt et al. (1997) examined the extent to which skin-mounted markers represented skeletal knee joint motion during running. Mean error as a percentage of the joint range of motion during running stance were 21%, 63%, and 70% for knee flexion/extension, internal/external rotation, and abduction/adduction, respectively, and were highly subject specific. Skin movement artifacts, tended to be very subject, segment, and task specific, and usually did not follow a consistent, predictable relation to bone-mounted marker motion.

2.3. Muscle Activation Patterns During Landing

Electromyography (EMG) is used to identify muscle activation patterns. These patterns can be used to identify how the lower extremity muscles provide coordinated movement during landing. There are several problems associated with reading EMG signals, however. The signal is dependant upon the muscle being measured (degree of superficiality), signal impedance (marker placement, skin, and hair), and subject body composition (electrodes on an obese subject will be further from the muscle than on a lean subject). These complications make EMG signal replication almost impossible.

McMahon et al., (1979) found leg stiffness, as defined in their paper, to decrease with increasing knee flexion, increase with increasing load on the leg, and is maximal when the knee is fully extended and rigidly locked. They also indicated that muscles can be preprogrammed to control joint motion prior to impact. It was suggested that future studies measure EMG to indicate the contribution of preparatory muscle action coupled with multi-joint motion just after initial ground contact on the influence of vertical impact peak force.

Santello et al. (1998) examined the relation between EMG timing and amplitude with vertical drop height during barefoot landings. They looked at soleus and tibialis anterior EMG and vertical ground reaction force data sampled at 2000 Hz. Six male subjects performed 10 drop jumps from each of five heights (0.2, 0.4, 0.6, 0.8 and 1.0 m) landing with both feet simultaneously onto a force plate. Mean vertical ground reaction force for the increasing drop heights were 3.9, 4.7, 5.6, 6.9, and 7.9 BW. Subjects demonstrated consistent co-contraction patterns before and after touchdown. Tibialis anterior EMG was the same across drop height, but the amplitude increased with increased drop height. Soleus preparatory EMG activity was shorter in duration for the 0.2 m drop height as compared with the other heights. The authors made a distinction between EMG activity associated with “takeoff” and that associated with “preparation for landing.” No reason for this distinction was given.

Kovacs et al. (1999) examined the influence of foot placement on lower extremity muscle activation patterns and the kinematics and kinetics of landing during drop jumping. Ten male students performed drop jumps from a 0.4 m high platform placed 1.0 m horizontally away from the force plate varying “foot placement” as heel-toe or fore foot. Three trials of each landing condition were performed and surface EMG activity of the gluteus maximus, vastus lateralis, and lateral head of the gastrocnemius muscles were recorded. Gluteus, vastus lateralis and plantar flexor muscle activity was similar between landing styles. The only exceptions were in the pre-contact phase where greater vastus lateralis activity occurred in the heel-toe landing condition while greater gastrocnemius activity was observed during the forefoot landing condition.

2.3.1. The Influence of Performance Feedback and Task

In a study by Sidaway et al. (1989), they proposed that visual feedback regulates muscle pre-activation prior to landing. This pre-activation allows the musculo-skeletal system to improve control of the body and land injury-free. They collected ground reaction forces and rectis femoris EMG data from 6 male subjects. Each subject performed drops by stepping from 0.72, 1.04, and 1.59 m heights landing with both feet simultaneously on a force plate. The subjects were instructed to watch the force plate as they landed. Mean preactivation interval times, the time from EMG onset to initial foot contact, for the rectus femoris were 60, 73, and 149 ms for the low, medium, and high drop heights respectively. The low and medium intervals were significantly lower than that for the higher height. The same authors (Sidaway et al., 1989) reported no preactivation in the biceps femoris. However, the vastus medialis and gastrocnemius demonstrated similar patterns to the rectus femoris. The patterns for each muscle seemed similar regardless of whether the subject stepped from a platform or dropped from a hanging position.

McNitt-Gray et al. (2001) examined 6 male gymnasts and the influence of body orientation at contact on mechanical joint loading. They also looked at the multijoint control of vertical ground reaction forces during various aerial maneuver landings. Each subject performed drop jumps, front and back saltos from a 0.72 m platform landing with both feet landing simultaneously on a force platform. A single successful trial was completed by each subject for each task onto doubled gymnastic mats (0.12 m thickness each) lying over a force plate. Kinematic (200 Hz), vertical ground reaction forces (800 Hz) and muscle activation patterns via EMG (1600 Hz) of seven lower extremity muscles (gluteus maximus, semitendinosis, biceps femoris, rectis femoris, vastsu medialis, medial gastrocnemius, and tibalis anterior) were measured. Knee resultant joint moments were in a flexion direction during front saltos and were significantly greater than those for drop landings. During all three landings tasks, knee and hip resultant joint moments opposed each other. Front salto and drop landings resulted in a knee flexion resultant joint moment that appeared to be countered by a hip extension resultant joint moment. During the stabilization phase, lower extremity resultant joint moments showed no significant differences between landing tasks. The authors suggested that the differences in muscle activation observed prior to contact were due to the gymnasts scaling their muscle activation patterns in preparation for the resultant joint moments incurred immediately after contact. Individual preferences, joint control issues, muscle moment arms and/or ability to produce force at given muscle lengths and rate of contraction may be the cause of the individual variability seen in muscle activation patterns and lower extremity resultant joint moments.

2.3.2. Gender Differences

Physiological differences have been identified between the genders, yet it remains unclear as to whether these differences are a direct result of gender (e.g. hormonal levels, growth and maturation cycle, reproductive organ development) or can be attributed to environmental or learning factors (e.g. physical activity level, education, opportunity). Males tend to be larger, stronger, and faster than females. They also tend to have fewer knee, specifically anterior cruciate ligament (ACL), injuries (Arendt et al., 1995). Little research has been done, however, to identify specific gender differences observed during landing from a jump/drop or impacts that may be predisposing females to knee injuries. The following two studies were reviewed because the activities included in their research closely resemble the athletic, quick start-stop type movements seen when athletes injure their ACL's.

Ives et al. (1993) examined potential gender differences in kinematics and EMG patterns during fast self-terminated elbow flexion movements. Ten males and ten females were instructed to move their right arms as fast as possible through a 90 degree flexion range of motion while held inside a cuff device affixed to a table top. Subjects were asked to stop as close as possible to a designated termination point. An electromagnet provided 0, 40, and 70% of maximum isometric flexion resistance. Ten trials were completed at each condition. Surface EMG was measured from biceps brachii and triceps brachii. The male subjects had faster movements and accelerations nearly double that of the females, under all conditions. EMG activation rise (especially in biceps) was greater in males compared with females, and males consistently demonstrated a more tightly coupled reciprocal activation of biceps and triceps. Both males and females tended to overshoot the termination target, but not differentially. These results suggest that females provided less rapid triceps activation in relation to biceps activation. Gender performance differences were attributed in part to neuromuscular coordination mechanisms, with males able to begin the braking process sooner. Results highlighted the importance of antagonist activity for both controlling braking and regulating acceleration.

Behm et al. (1996) examined the effect of joint angular velocity on agonist and antagonist muscle activation during isometric and concentric dorsi-flexion activity. Five males and females performed maximum isometric and isokinetic dorsi-flexion actions on a Cybex II isokinetic dynamometer. The best trial (e.g. greatest generated moment) at was selected for analysis. EMG of tibalis anterior, soleus, and lateral gastrocnemius were sampled during all trials. Males produced greater absolute ankle moment (22.2 ± 7.6 Nm) compared with females (17.4 ± 5.2 Nm), with no gender differences noted in moment to body mass ratio. Males demonstrated a greater ratio of soleus to tibialis anterior compared with gastrocnemius to tibialis anterior activation. The opposite was true for the females.

2.4 Summary

This literature review has illustrated that the assumption that the body segments are rigid is not necessarily accurate. This is especially true when landing from a drop, when high impact forces are experienced. The relevant literature regarding traditional kinematic and kinetic analyses of landings and the downside of common assumptions were presented along with typical ground reaction force values, resultant joint moments and muscle activation patterns observed during various landing impacts, as well as during relevant single joint tasks. Analysis of EMG data suggests that muscles can be preprogrammed to complete a landing task. This preprogramming is dependant on the task and the skill of the participant. Studies have reported general trends in gender differences during landing, however there is a apparent lack of evidence regarding specific kinetic and kinematic gender differences.

CHAPTER 3

METHODS

3.1. Introduction

To examine the difference in landing performance the two groups of subjects were asked to perform landings from a nominal drop height of 0.35 m. The following sections describe the experimental protocol, the methods used to collect and process data, the key variables determined, and how they were statistically compared.

3.2 Experimental Protocol

Eight varsity level female gymnasts and eight varsity level female swimmers, aged between 19 and 23 years of age, participated in this study. The two subject groups had the following characteristics: gymnasts - height 1.62 ± 0.3 m, mass 65.26 ± 5.3 kg; swimmers - height 1.71 ± 0.5 m, mass 59.01 ± 3.0 kg. Each subject reported no incidence of current lower extremity injury, and volunteered for participation in this study. The subjects all provided informed consent, and all procedures were approved by The Pennsylvania State University Institutional Review Board. Each subject performed ten two-footed landings onto a force plate from a nominal drop height of 0.35 meters. Their actual drop heights could vary from the nominal height if a subject lowered or raised their center of mass as they left the 0.35 meter high platform.

Subjects were asked to step off the 0.35 m high platform and land in their preferred style, coming to an eventual stop in an upright position. During these tasks they could use their arms in any way they chose to control their landing. For safety the subjects wore their preferred footwear, which were sneakers.

3.3 Motion Analysis

Positional data of markers placed on key anatomical landmarks were recorded during the landings in two dimensions utilizing a Qualisys Pro-Reflex motion analysis system. Position data were sampled at 240 Hz, and had a measurement accuracy of 0.01 % of the field of view. The optical axis of the camera was perpendicular to the plane of motion with the camera positioned 6 meters away from the force plate.

To calculate kinetic and kinematic variables, the body was considered as a series of rigid bodies composed of four segments (foot, shank, thigh and upper body) linked by frictionless hinge joints. Eight spherical retro-reflective markers were affixed on the right side of each subject, in order to identify these segments, at the following anatomical landmarks: shoulder (acromion process), hip (greater trochanter), ankle (lateral malleolus), heel (lateral posterior calcaneous), and toe (5^{th} metatarsal phalanx). The model used in this study is illustrated in figure 3.1, as well as the definitions of segments and joint angles. The upright stance at the end of the analyzed movement was used to define the reference angles: ankle - 90 degrees, knee - 180 degrees, and hip - 180 degrees.

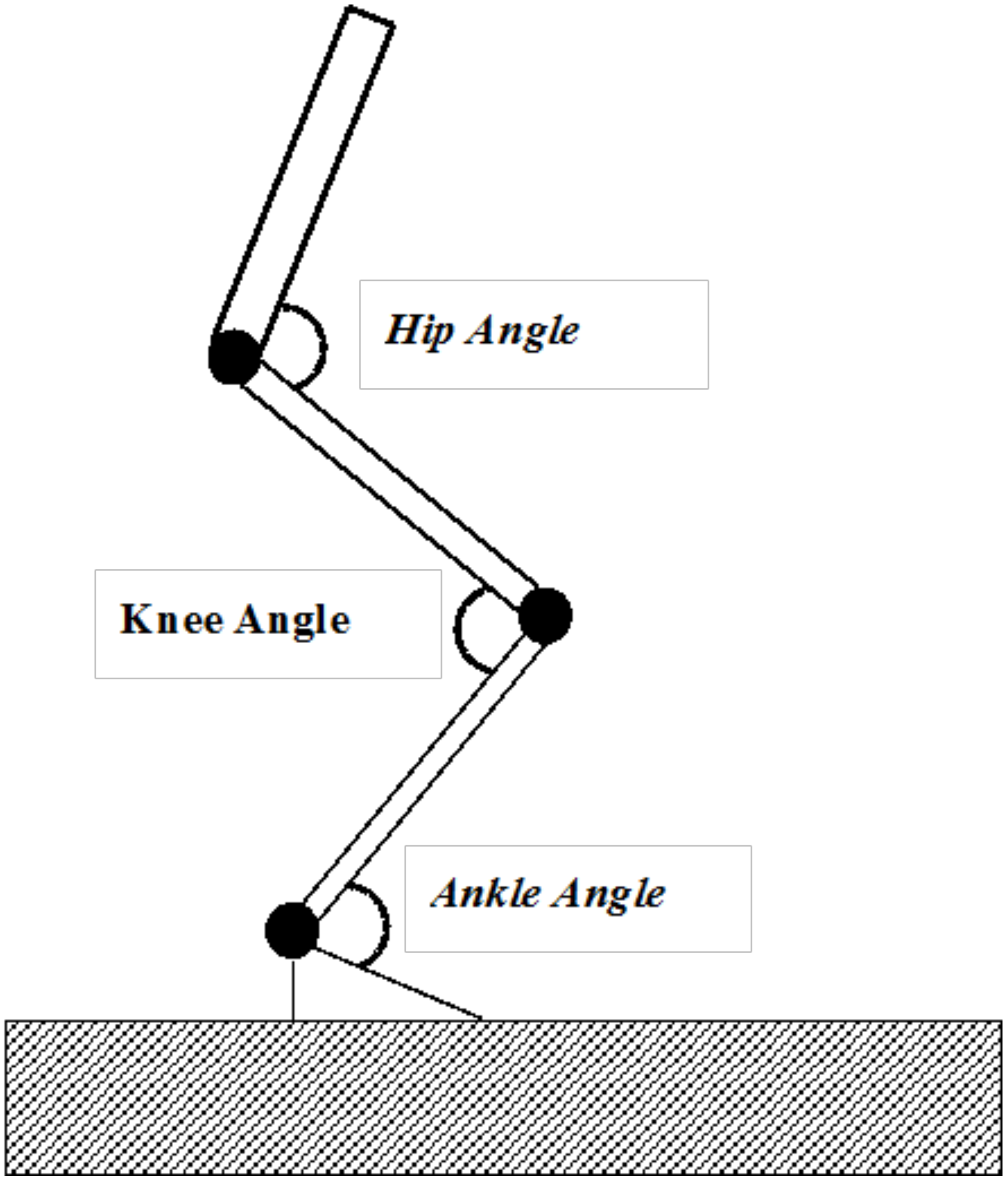

Figure 3.1. Joint angle definitions.

The coordinates for each body landmark were filtered using a cut-off frequency, which was estimated by exploiting the properties of the autocorrelation function of white noise. The cut-off frequencies were selected using the technique presented in Challis (1999). The procedure systematically varied the cut-off frequency of a Butterworth filter until the signal representing the difference between the filtered and unfiltered data was the best approximation to white noise as assessed using the autocorrelation function (Challis, 1999). A second order Butterworth filter was used, applied in both forward and reverse direction to remove phase lag, effectively doubling the order of the filter. Problems at data boundaries have been highlighted for the Butterworth filter (Smith, 1989), and to circumvent these problems the data sets in this study were padded, by reflecting the first 30% of the data at the beginning of each data set, and the last 30% at the end. Derivatives were computed using first order finite difference equations.

3.4 Ground Reaction Force Data

Ground reaction forces were recorded at 240 Hz during the landing trials using a Bertec force plate (#N50601, Type 4080s). Force plate sampling was electronically synchronized via a 240 Hz synchronizing pulse from the Pro-Reflex system. Vertical and horizontal forces and the coordinates of the center of pressure were measured. An analog low-pass filter filtered the force plate data with a cutoff frequency of 120 Hz. This cutoff frequency is less than the natural frequencies of the plate in situ. The filter effectively removed the high frequency noise in the signal, leaving the frequency components associated with the landing impact intact.

The landing period, determined from the vertical ground reaction force time profiles, was defined from the instant of contact to the termination of movement (determined when vertical ground reaction force levels reflected body weight). Integration with respect to time of the vertical ground reaction force for this period elicited impulse, from which vertical landing velocity and then the actual drop height were computed. Further integration provided the displacement of the whole body center of mass during the landing.

The variables of landing velocity, peak vertical ground reaction force magnitude, time of peak vertical ground reaction force, time at which the subject's center of mass was at it's lowest vertical position, and time of movement termination were extracted from the force plate data.

3.5 Determination of Body Segment Inertial Properties

The inertial parameters of the thigh, shank and foot were determined. Segment boundaries for these segments were defined as specified in the work of Dempster (1955) and identified by the measurer. The subjects were measured using an anthropometric measuring tape and all measurements were made to the nearest millimeter. Measurements were performed on both lower limbs of each subject with the subject standing upright. For each segment, a system of axes was defined with its origin at the center of mass of the segment. The axes were aligned with approximate body axes: sagital (*x*), frontal (*y*) and longitudinal (*z*).

The inertial properties of the thigh and shank were determined by modeling them both as a series of two truncated cones. These segments were divided into equal halves and measures taken to model each half as a truncated cone: a perimeter at the top, one at the bottom and the distance in between the perimeters. The foot was modeled as a series of two stadium solids (Yeadon, 1990). The density values for these solids were taken from the data of Clauser et al. (1969) and were assumed to be uniform throughout a given segment.

3.6 Resultant Joint Moments

Resultant joint moments at each point were computed for the ankle, knee, and hip during the jumping period by combining the kinematic, ground reaction force and anthropometric data for each subject (Winter, 1990). Equations 3.1, 3.2, and 3.3 illustrate how the joint reaction forces and resultant joint moments were calculated. Resultant joint moments were computed in a bottom to top fashion.

$$R_{xp} - R_{xd} = m \cdot a_x \qquad [3.1]$$

$$R_{yp} - R_{yd} - m \cdot g = m \cdot a_y \qquad [3.2],$$

Where,

R – joint reaction force

m – segment mass

a – acceleration of segments center of mass

g – gravitational acceleration

x,y – indicate horizontal and vertical directions, respectively

p,d – indicate proximal and distal ends of a segment, respecively

$$\Sigma M = I \cdot \alpha = M_d + M_p + M_{jf} \qquad [3.3]$$

Where,

ΣM – sum of all the moments acting about the segments center of mass

I – moment of inertia of the segment about its center of mass

α – angular acceleration of the segment mass

M_d – moment acting at the distal joint of the segment

M_p – moment acting at the proximal joint of the segment

M_{jf} – is the moment due to both joints reaction forces

Resultant joint moments are assumed to represent forces from the muscles crossing that joint, with passive effects of other tissues being minimal (Andrews, 1982). It is assumed that the segments in the model were rigid and connected by frictionless hinge joints, and motion was assumed to occur in the sagittal plane. For the ankle, knee, and hip joints extensor moments were designated positive. The moments for each subject were normalized by dividing by the subject's body weight.

3.7 Statistical Analysis

Means and standard deviations of all variables across subject, and subject group were computed. A repeated measures analysis of variance was performed in MINITAB (version 12.1, Minitab Inc.) on the trial and subject group data. Homogeneity of variance was confirmed for the data prior to performing the general linear model using a Bartlett test. A significance level of 0.05 was selected.

Comparisons were made of the inertial properties between the two groups. To remove the influence of one group having different inertial properties compared with the other because they were larger these data were normalized. The segment masses were normalized with respect to subject mass, the segment center of mass locations were normalized with respect segment length, and segmental transverse moment of inertia with respect to the product of subject mass and the square of subject height.

Ground reaction force data were normalized with respect to subject body weight. Resultant joint moments were normalized with respect to subject mass and segment height (Hof, 1996).

3.8 Summary of Analyzed Variables

Table 3.1 summarizes all the analyzed variables and their units, except for the anthropometric variables. Note that all timings are relative for time of initial contact with the force place which is assumed to occur at zero seconds. Descent is defined as the part of the movement when the center of mass is moving downwards, and ascent when the center of mass is moving up. The majority of the analysis was for the landing phase only.

Table 3.1 – summary of all analyzed variables, their source, and their units.

Variable	Units
Landing Velocity	m/s
Drop height	m
Maximum center of mass displacement	m
Peak vertical ground reaction force	N
Time of peak vertical ground reaction force	s
Time center of mass is at bottom of trajectory	s
Landing Duration	s
Ankle, knee, and hip angles at impact	deg
Ankle, knee, and hip angles at peak VGRF	deg
Ankle, knee, and hip angles at center of mass lowest point	deg
Ankle, knee, and hip ranges of motion throughout landing	deg
Peak ankle, knee, and hip angular velocities during descent	rad/s
Time of peak ankle, knee, and hip angular velocities	s
Peak ankle, knee, and hip resultant joint moment during descent	N.m
Time of peak ankle, knee, and hip resultant joint moment during descent	s
Peak ankle, knee, and hip angular velocities during ascent	rad/s
Time of peak ankle, knee, and hip angular velocities during ascent	s
Peak ankle, knee, and hip resultant joint moment during ascent	N.m
Time of peak ankle, knee, and hip resultant joint moment during ascent	s

3.9 Summary

This Chapter has described the methods used to analyze the landings from the drops performed by the two subject groups: gymnasts and swimmers. The analysis has combined force plate and motion analysis data to provide variables which describe these landings. The results and statistical comparisons of these data will be reported in the next Chapter.

CHAPTER 4

RESULTS

4.1 Introduction

This chapter reports the results from the landings from the drops of the two groups, swimmers and gymnasts. The following sections report the subject inertial properties, center of mass motion, ground reaction forces, joint kinematics, and joint kinetics of the two groups and the statistical comparison of those values.

4.2 Subject Inertial Properties

The height of the gymnasts (1.62 m ± 0.03) was statistically smaller than that of the swimmers (1.71 m ± 0.05). However, the overall mass of the gymnasts (65.26 kg ± 5.3) was statistically greater than that of the swimmers (59.01 kg ± 3.0). The difference between the body mass index of the gymnasts (22.53 kgm^{-2} ± 1.8) and the swimmers (22.23 kgm^{-2} ± 1.6) was not statistically significant.

Both the raw and normalized values for the gymnast's and swimmer's mass, center of mass, moment of inertia, and length for each the thigh, shank, and foot were compared using a one-tailed t-test (Table 4.1). Only the non-normalized mass of the thigh and foot were found to be statistically significant between the gymnasts and swimmers.

Table 4.1 - Means and standard deviations of thigh, shank and foot segment inertial parameters for the two subject groups.

Segment	Swimmers		Gymnasts	
	Actual	Normalized	Actual	Normalized
THIGH				
Mass (kg)	8.289 ± .943	0.141 ± .017	7.833 ± 1.213	0.120 ± .017
Length (m)	0.434 ± .0298	0.254 ± .016	0.411 ± .046	0.254 ± .030
C. of M. (m)	0.186 ± .011	42.91 ± 0.56	0.175 ± .020	42.63 ± 0.89
I_T (kg.m^2)	0.134±0.0321	7.71 ± 1.45	0.116 ± .0399	6.76 ± 2.22
SHANK				
Mass (kg)	3.095 ± .470	0.053 ± .010	3.264 ± .560	0.050 ± .010
Length (m)	0.387 ± .033	0.226 ± .019	0.385 ± .026	0.238 ± .013
C. of M. (m)	0.166 ± .014	42.92 ± 1.03	0.170 ± .013	44.11 ± 1.22
I_T (kg.m^2)	0.037 ± 0.010	2.15 ± 0.60	0.0383 ± .0094	2.25 ± 0.56
FOOT				
Mass (kg)	0.926 ± .258	0.016 ± .004	0.687 ± .299	0.010 ± .004
Length (m)	0.216 ± .027	0.126 ± .014	0.176 ± .066	0.109 ± .041
C. of M. (m)	0.089 ± .006	41.72 ± 3.31	0.074 ± .028	42.02 ± 3.57
I_T (kg.m^2)	0.0039±0.002	2.20 ± 0.97	0.0240 ± .0014	1.40 ± 0.83

Note - The normalized values for center of mass (C. of M.) is the percentage of the segment length the center of mass is located from the proximal joint. The actual values are in meters. The moment of inertia (I_T) is the transverse moment of inertia about the segments center of mass. These normalized moment of inertia values have been multiplied by 10,000.

4.3 Center of Mass Motion

The landing velocity, maximum of center of mass displacement, time center of mass at bottom of trajectory, landing duration, and drop height for both the gymnasts and swimmers were compared using a repeat measures ANOVA (Table 4.2). Only the maximum center of mass displacement was found to be statistically significantly different between the gymnasts and swimmers, with the swimmers having greater motion. For none of these metrics was there any statistically significant effect caused by trial order.

Table 4.2 – Means (±standard deviation) of metrics of center of mass motion during the landings from the drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Landing Velocity (m/s)	-2.6 ± 0.23	-2.5 ± 0.23
Maximum center of mass displacement (m)*	0.17 ± 0.06	0.29 ± 0.08
Time center of mass at bottom of trajectory (s)	0.238 ± 0.064	0.321 ± 0.091
Landing Duration (s)	1.210 ± 0.397	1.333 ± 0.305
Drop height (m)	0.34 ± 0.060.	0.31 ± 0.05

Note - * indicates a statistically significant difference between the groups $p < 0.05$.

4.4 Ground Reaction Forces

The peak vertical ground reaction force, normalized peak vertical ground reaction force, time of peak vertical ground reaction force, loading rate, and normalized loading rate were compared using a repeat measures ANOVA (Table 4.3). All of these metrics were found to be statistically significantly different between the gymnasts and swimmers, with each being statistically significantly greater for the gymnasts, except for time of peak vertical ground reaction force. An example of this can be seen in figure 4.1. Trial order also caused a statistically significant difference in the time to peak vertical ground reaction force between the gymnasts and swimmers, and there were trends toward statistical significance in loading rate and normalized loading rate, as well.

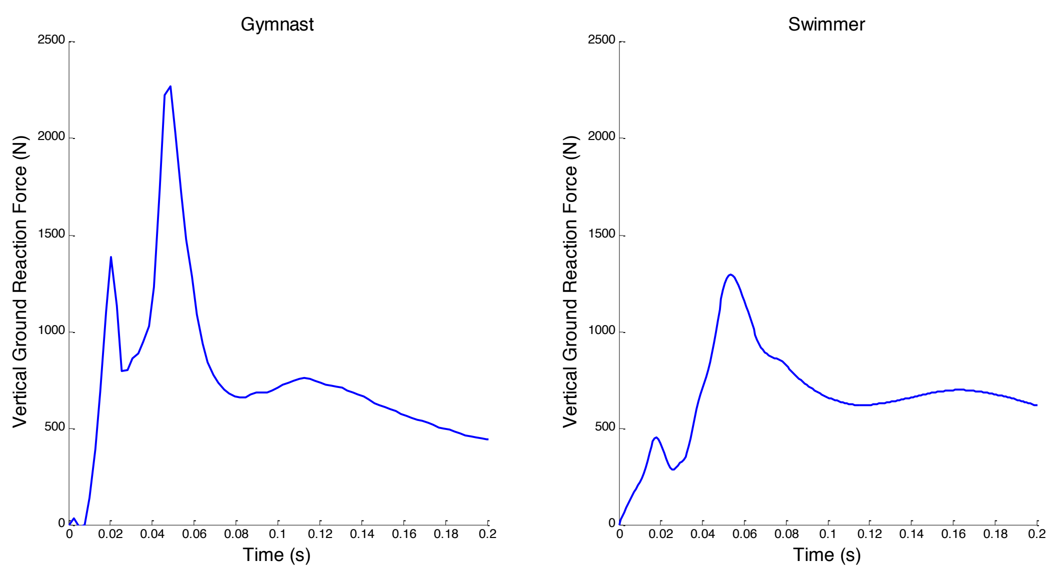

Figure 4.1. Vertical Ground Reaction Force for the drop landings for a representative gymnast and representative swimmer.

Table 4.3 – Means (±standard deviation) of metrics of the vertical ground reaction force during the landings from the drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Peak vertical ground reaction force (N)*	1892 ± 509	1121 ± 288
Normalized Peak vertical ground reaction force (N/Bw)*	2.97 ± 0.82	1.94 ± 0.52
Time of peak vertical ground reaction force (s)*^	0.0426 ± 0.007	0.0512 ± 0.0084
Loading rate ($N.s^{-1}$)*	47503 ± 22546	22767 ± 8456
Normalized Loading rate ($(N/Bw).s^{-1}$)*	75.0 ± 37.4	39.5 ± 14.5

Note - * indicates significant difference between groups ($p < 0.05$), and ^ indicates a significant effect of trial ($p < 0.05$).

4.5 Joint Kinematics

The kinematics for the joint angle at impact, at peak VGRF, at center of mass's lowest position, and range of motion at the ankle, knee and hip were compared between the two groups using a repeat measures ANOVA (Table 4.4). The joint angle at peak VGRF at the knee, and the joint angle at center of mass's lowest position at the knee and hip were all statistically significantly greater for the gymnasts. An example of this can be seen in Figure 4.2. The joint angle range of motion at the knee and hip were statistically significantly greater for the swimmers. There were trends toward statistical significance in the joint angle at peak VGRF at the hip and the joint angle at center of mass's lowest position at the ankle for the gymnasts, as well as in the joint angle range of motion at the ankle for the swimmers. Trial order had a statistically significant effect on the joint angle at the ankle and hip for the gymnasts at the center of mass's lowest position. There were also trends towards statistical significance in the joint angle at peak VGRF at the hip for the gymnasts.

The peak joint angular velocities during the descent from the landings from the drop for the gymnasts and swimmers were compared using a repeat measures ANOVA (Table 4.5). The peak joint angular velocity at the knee was statistically significantly greater for the swimmers compared with the gymnasts. An example of this can be seen in Figure 4.3. Trial order also had a statistically significantly effect at the hip for the gymnasts.

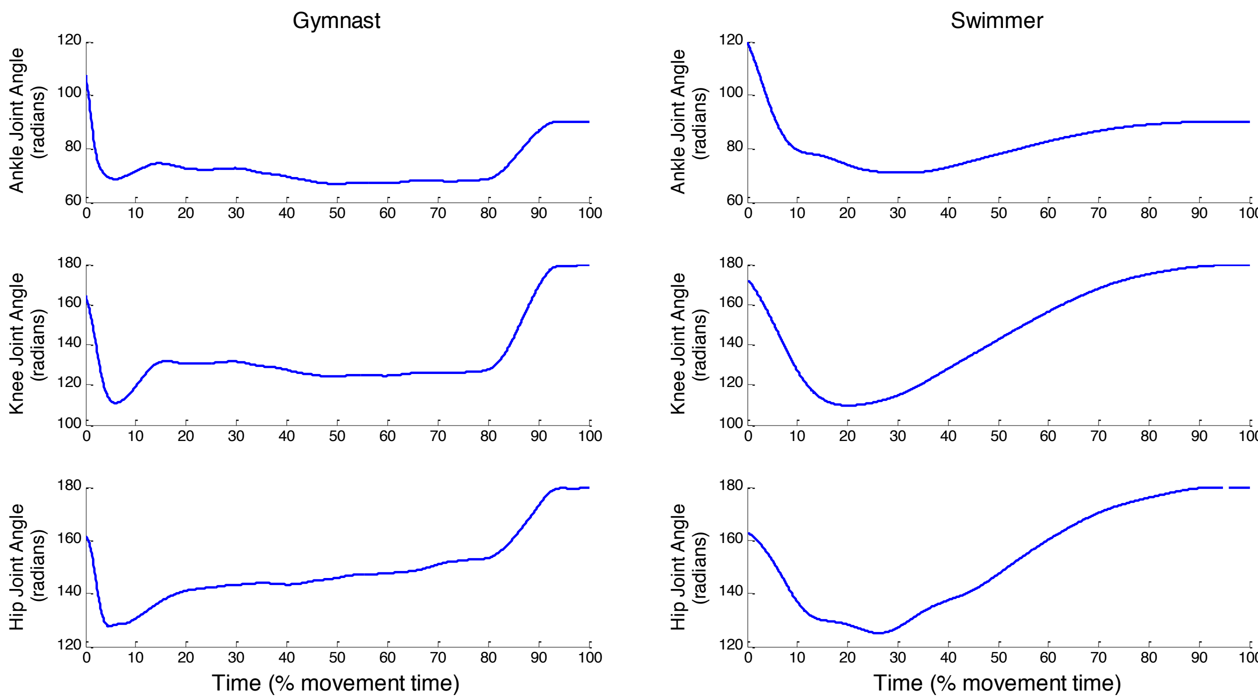

Figure 4.2. Lower limb joint angles during the drop landings for a representative gymnast and swimmer.

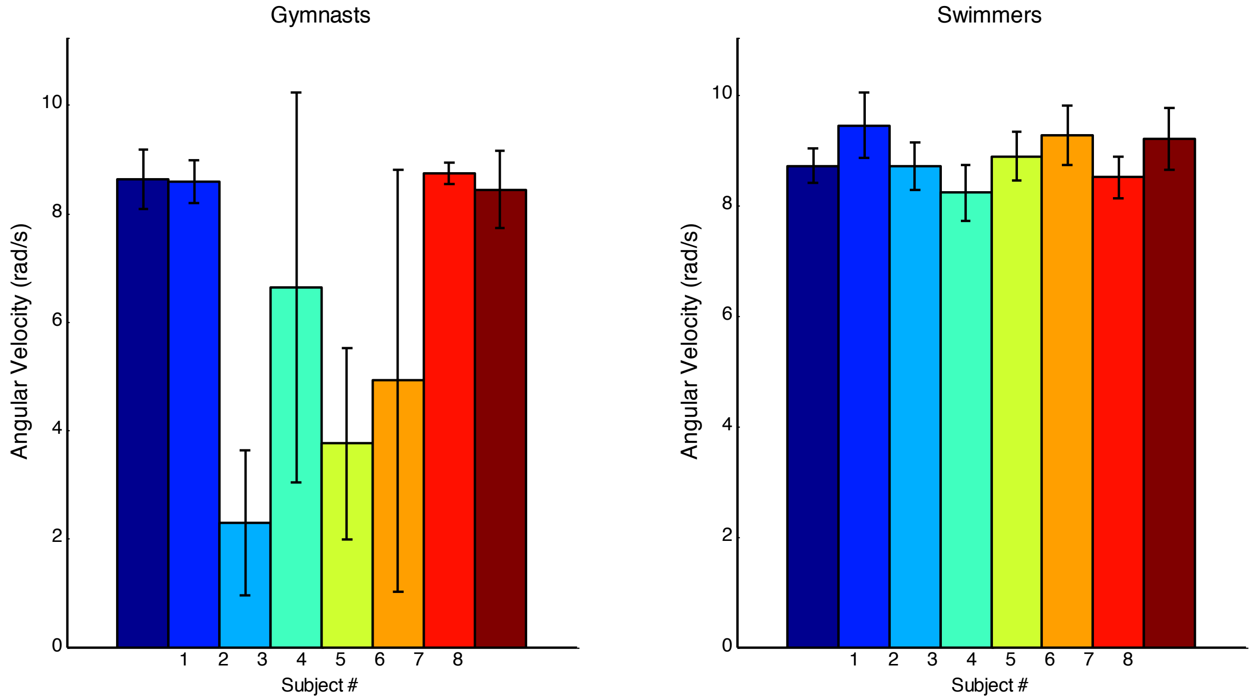

Figure 4.3. The mean and standard deviation of the peak angular velocity for the knee joint for each gymnast and swimmer across their 10 landing trials.

Table 4.4 – Means (±standard deviation) of joint angles during key events at key times during the landings from the drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Joint Angles at Impact		
Ankle (degrees)	102.2 ± 16.4	110.9 ± 8.2
Knee (degrees)	166.7 ± 13.2	162.2 ± 8.1
Hip (degrees)	150.3 ± 26.2	157.3 ± 10.1
Joint Angles at peak VGRF		
Ankle (degrees)	88.7 ± 7.1	85.9 ± 6.4
Knee (degrees)*	152.6 ± 14.2	138.3 ± 8.7
Hip (degrees)	149.5 ± 13.9	142.2 ± 11.2
Joint Angles at Center of Mass's Lowest Position		
Ankle (degrees)^	76.1 ± 10.4	67.1 ± 7.72
Knee (degrees)*	123.8 ± 20.3	93.8 ± 14.0
Hip (degrees)*^	123.3 ± 25.1	88.1 ± 24.4
Joint Angles Range of Motion		
Ankle (degrees)	31.9 ± 18.7	44.7 ± 9.6
Knee (degrees)*	57.8 ± 19.5	87.5 ± 13.8
Hip (degrees)*	58.2 ± 19.1	88.6 ± 13.2

Note - * indicates significant difference between groups ($p < 0.05$), and ^ indicates a significant effect of trial ($p < 0.05$).

Table 4.5 – Means (±standard deviation) of peak joint angular velocities during the descent for the landings from the drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Ankle (radians/second)	5.1 ± 3.6	7.4 ± 1.7
Knee (radians /second)*	6.6 ± 3.1	8.9 ± 0.6
Hip (radians /second)^	8.4 ± 5.3	7.1 ± 1.4

Note - * indicates significant difference between groups ($p < 0.05$), and ^ indicates a significant effect of trial ($p < 0.05$).

4.6 Joint Kinetics

The normalized peak resultant joint moments (RJM), and the time of those normalized peak RJM, during the descent and ascent phase of the landing from a drop at the ankle, knee, and hip were compared between the two groups using a repeat measures ANOVA (Table 4.4, Figure 4.4). During the descent, the time of knee and hip peak RJM were statistically significantly greater for the swimmers compared to the gymnasts. An example of this, for the knee, can be seen in figure 4.5. During the ascent, there were no statistically significant values between the two groups.

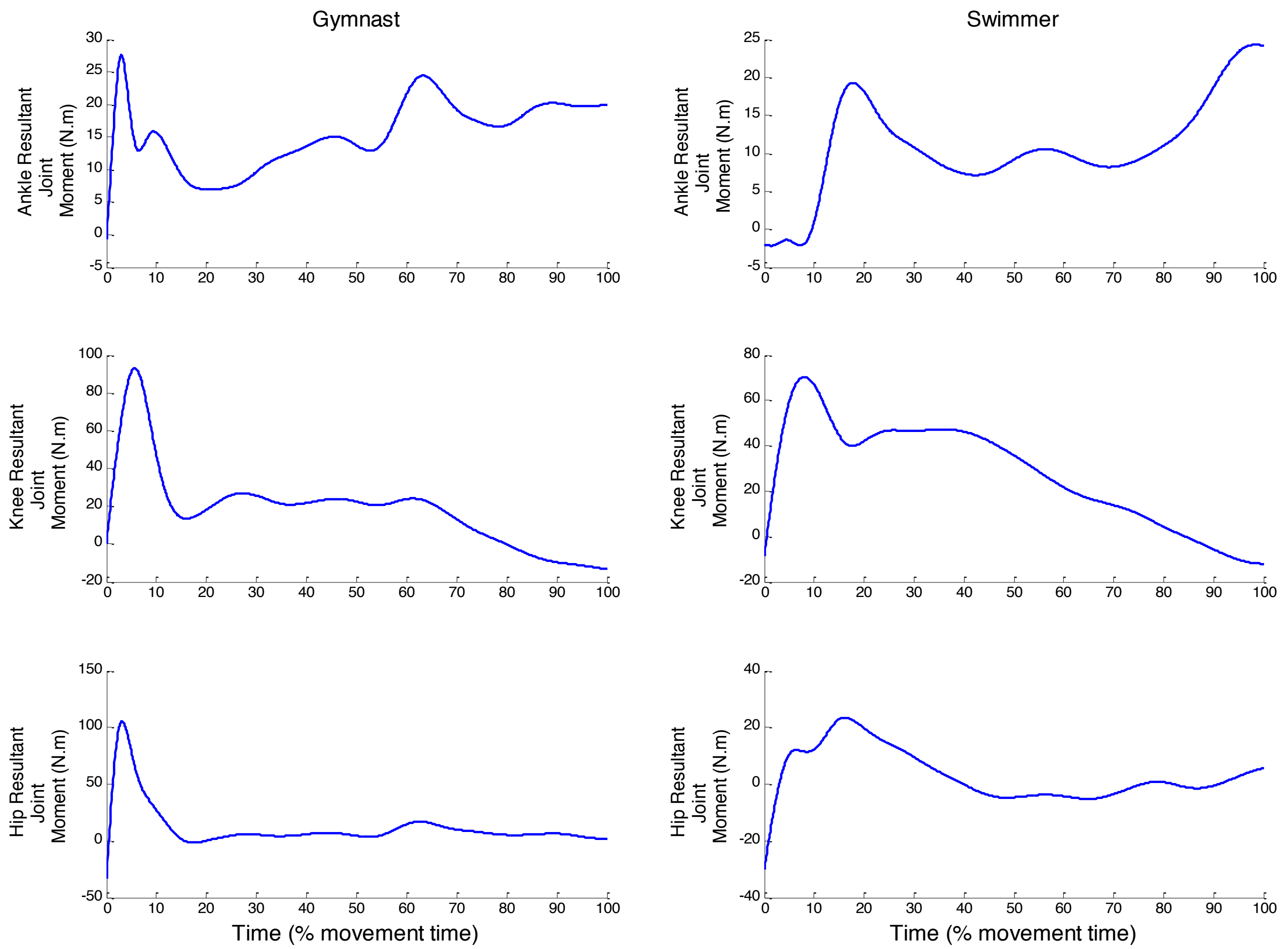

Figure 4.4. Lower limb resultant joint moments during the drop landings for a representative gymnast and swimmer for the ankle, knee, and hip joints.

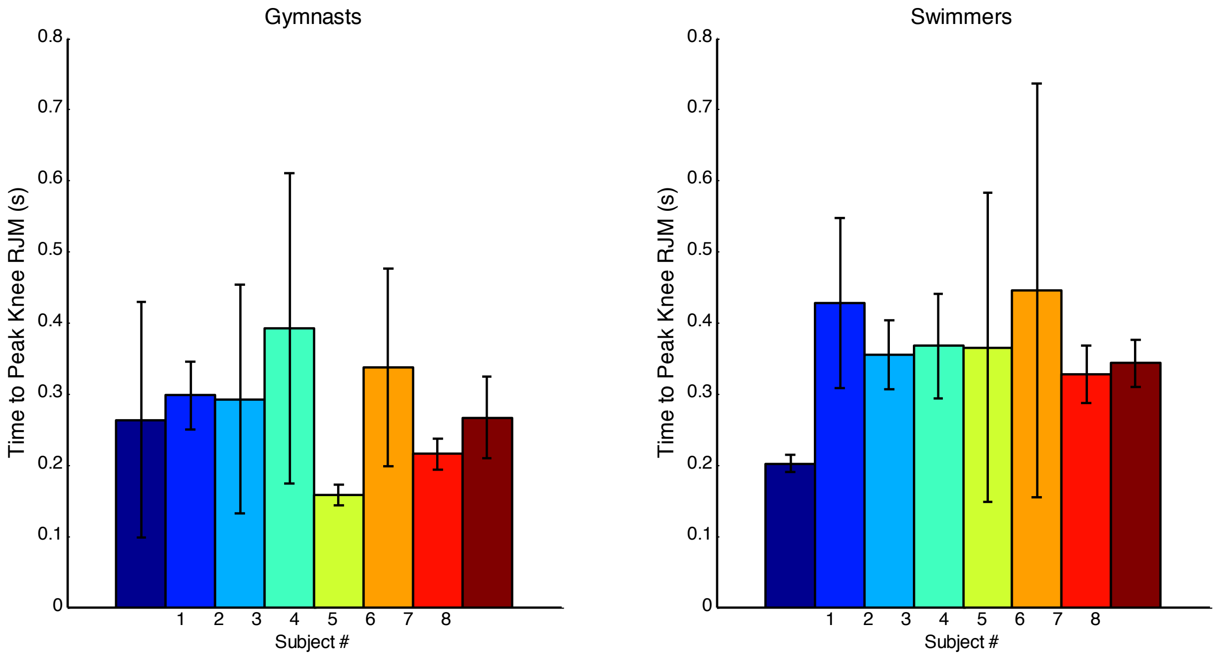

Figure 4.5. The mean and standard deviation of the time to peak knee resultant joint moment for each gymnast and swimmer across their 10 landing trials.

Table 4.6 – Means (±standard deviation) of the peak resultant joint moments (RJM) during the descent phase of the landing from a drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Normalized Ankle Peak RJM (N/(Bw.Ht) %)	3.4 ± 2.9	3.4 ± 2.1
Time of Ankle Peak RJM (s)	0.076 ± 0.067	0.119 ± 0.117
Normalized Knee Peak RJM (N/(Bw.Ht) %)	5.2 ± 3.6	6.7 ± 4.3
Time of Knee Peak RJM (s) *	0.115 ± 0.055	0.173 ± 0.079
Normalized Hip Peak RJM (N/(Bw.Ht) %)	4.4 ± 2.1	4.4 ± 3.6
Time of Hip Peak RJM (s)	0.060 ± 0.028	0.102 ± 0.083

Note - * indicates significant difference between groups ($p < 0.05$), and ^ indicates a significant effect of trial ($p < 0.05$).

Table 4.7 – Means (±standard deviation) of the peak resultant joint moments (RJM) during the ascent phase of the landing from a drop for the gymnasts and swimmers.

	Gymnasts	**Swimmers**
Normalized Ankle Peak RJM (N/(Bw.Ht) %)	5.5 ± 4.0	4.3 ± 4.2
Time of Ankle Peak RJM (s)	0.662 ± 0.493	0.940 ± 0.403
Normalized Knee Peak RJM (N/(Bw.Ht) %)	8.5 ± 3.8	8.6 ± 4.7
Time of Knee Peak RJM (s)	0.326 ± 0.295	0.370 ± 0.188
Normalized Hip Peak RJM (N/(Bw.Ht) %)	12.1 ± 4.3	8.3 ± 5.4
Time of Hip Peak RJM (s)	0.447 ± 0.387	0.494 ± 0.302

4.7 Summary

This chapter has reported the results from the landings from the drops of the two groups. Several variables were compared statistically to find significance between the two groups or based on trial order. The analysis of this data will be reported in the next chapter.

CHAPTER 5

DISCUSSION

5.1 Introduction

In the following sections the results of this study are summarized, the following section discusses the results. The next two sections outline the primary limitations of the study and future studies. The Chapter ends with a conclusion.

5.2 Summary of Findings

The aims of this study were to analyze the inertial properties, center of mass motion, ground reaction forces, joint kinematics, and joint kinetics of two groups, gymnasts and swimmers, as they landed from a drop.

Not surprisingly, the gymnasts were statistically shorter and heavier than the swimmers, and for almost all body segments, the center of mass location, moment of inertia, and the length for each thigh, shank, and foot were not statistically significantly different between the two groups when normalized for body size. These differences were anticipated when selecting the two groups, and were not seen as important as the differences in their experience with landing from a drop.

Almost all of the metrics for ground reaction forces were statistically significant different between the groups, with the gymnasts having the greater values. This was actually surprising. It was assumed that since the gymnasts were more experienced at landings than the swimmers that they would be better at dissipating their ground reaction forces. What seemed to occur was that since the landing was from such a small height, compared to what they are used to, and they are required to "stick" their landings in competition, they did not care to minimize their ground reaction forces during the trials. As a point of comparison the higher bar on the uneven bars used by gymnasts is 2.50 m from the ground, so they would drop a much greater distance when dismounting from this apparatus than used in this study (0.35 m).

Only the maximum center of mass displacement was found to be statistically significantly different between the gymnasts and swimmers, with the swimmers having greater motion. This greater range of motion indicates the swimmers attempt to reduce the impact forces while landing from the drop, by arresting the momentum from landing over a longer period of time.

The joint angle at peak VGRF at the knee, and the joint angle at the center of mass's lowest position at the knee and hip were all statistically significantly greater for the gymnasts. This indicates a more upright posture at landing, supporting the idea that they did not try to mitigate their ground reaction forces; they performed stiffer landings (Zatsiorsky & Prilutski, 1987). The joint angle range of motion at the knee and hip and the peak joint angular velocity at the knee was statistically significantly greater for the swimmers. This also supports the idea that the swimmers used motion at the hip and knee to reduce their impacts at landing.

During the descent, the time of knee and hip peak RJM were statistically significantly greater for the swimmers compared with the gymnasts, again confirming that the swimmers used more knee and hip range of motion to reduce the forces associated with the landings.

5.3 Results Discussion

Comparing this study to published research highlights some interesting differences, similarities, and potential explanations. McNitt-Gray (1991) found similar results to the current study. In her study, the gymnasts had slightly greater knee and ankle extensions at contact versus the recreational athletes. The recreational athletes would be comparable to the swimmers in the current study. The recreational athletes also utilized greater hip flexion when landing from the higher heights and flexed the hip joint less from the low height and more from the higher height than the gymnasts.

McNitt-Gray et al. (1993) attempted to determine if female gymnasts would vary their self-selected landing strategies when landing on mats of the same thickness, but varied composition, and from varied heights. The results of the study showed that self-selected landing strategies of female gymnasts did differ under varying drop heights and mat compositions. As drop height increased, the degree of joint flexion, rate of joint flexion, impact peak magnitude, and landing phase duration all tended to increase. The stiffer mat also tended to elicit larger degrees of knee flexion. The differences in joint flexion and rate of joint flexion were smaller between different mat compositions as opposed to the different drop heights. This shows that the gymnasts in their study varied their landing strategies differently when adjusting for differences in height versus differences in mat composition. If drop height or landing area composition had been varied in the present study, different results may have been found, but the requirement of distal forces for the computation of resultant joint moments required a force plate providing a stiff landing surface.

Mizrahi et al. (1982) examined the influences of body position, range of flexion of the lower extremity joints, softness of the ground, and the use of a ground roll immediately on impact during landing. Their results emphasize the role of muscle activation and joint movements in reducing peak forces during landings, similar to the current study. Simultaneous use of all joints is probably the basis for reducing peak forces at impact, a strategy more evident in the kinematics of the swimmers than the gymnasts in this study.

In DeVita et al. (1992), similar to the swimmers in the current study, soft landings resulted from a larger range of motion at each joint when compared with the stiff landings, like the gymnasts. They also found the stiff landings produced greater hip and knee resultant joint moments during the descent phase and had greater vertical ground reaction forces. This corresponds with the gymnast's data in the current study.

5.4 Study Limitations

A major limitation of this study was that the motion analysis was conducted for one plane of motion only. The comparison of the three-dimensional moments at the knee joint indicate that varus moments may be key indicators of the reasons why males and females have different knee injury rates (Hewett et al., 2005); this study did not examine these moments but a future study could. The present study only included only female subjects, but including males may have provided more insight into the females landing strategies by providing a point of comparison.

The subjects stepped off a platform and dropped onto a force plate. The two groups used different strategies for this, and as a consequence the swimmers actually dropped from a lower height than the gymnasts. Other studies have used a similar approach (e.g., DeVita et al., 1992; McNitt et al., 1993; Mizrahi et al., 1982). To have better experimental control over drop height, and therefore the task to be performed, subjects could drop from a trapeze.

Only one style of landing was examined, that is landing initially on the balls of the feet. There is evidence that in dynamic landings that injury rates are higher when the landing is on the heel (e.g., Steele, 1990). Presumably depending on how well they control their rotation gymnasts will on occasions land on their heels but this would be less common for swimmers. Gymnasts are also used to landing in bare feet which is likely not the case for the swimmers. It might have revealed more about their landing strategies to have both groups perform landings barefoot and shod.

5.5 Directions for Future Research

After reviewing the results and limitations of the current study, potential directions for future research are evident. While keeping a similar research design, future research could examine male subjects, perhaps also swimmers and gymnasts, to see if the same pattern of results emerges.

Resultant joint moments only give an indication of the net muscular activity, not what the individual muscles are doing. Therefore a future study could include the collection of muscle electrical activity via electromyography.

Such an analysis may provide additional insight into the mechanisms at play when landing from a drop.

There is evidence that landing strategies change in women with maturity (Quatman et al., 2006), therefore it would be valuable to compare the kinematics and kinetics of landing for both younger and older athletes. Indeed it would be interesting to track the two populations examined in this study for a range of ages.

Finally, it would be interesting to examine other populations of athletes to see their kinematics and kinetics in landing from a jump, for example volleyball athletes, and golfers.

5.6 Conclusions

After reviewing the results of the study, it was interesting to find that the gymnasts made less of an effort to dissipate the forces of the landing task then did the swimmers. The present study did have some limitations, primarily the use of motion analysis in one plane of motion, and future studies would benefit from the inclusion of males and older females.

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