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# Peak Torques of Lower Extremity Muscles Does Not Have Influence on Rate of Loading During Single Leg Drop-Landing

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**Abstract:** Shock absorption and reducing the Rate of Loading (ROL) during daily activities are considered as a reason for reducing the rate of injuries. Since, skeletal muscles utilize as shock absorber during weight bearing tasks, the purpose of this study was to examine the relationship between the peak torques of lower extremity muscles with the ROL during single leg drop landing. 33 healthy male students (23±2 years and 73±3 kg) from department of physical education and sport sciences volunteered in this study. Peak torque of lower extremity muscles and the Ground Reaction Forces (GRF) were measured by the Biodex system and force plate (AMTI), respectively. The Pearson correlation used to analyses the data at the $P \leq 0.05$ level. The mean torque of the quadriceps, hamstring, plantar flexors, dorsi flexors, invertors and evertors muscles were 2.70, 1.75, 0.77, 0.43, 0.43, 0.40 N.m respectively and the mean ROL was 519.60 N/ms. Significant correlations wasn't seen between peak torques of lower extremity muscles with ROL. Due to results of this study it seems that peak torques of quadriceps, hamstring, plantar flexors, dorsiflexors, invertors and evertors muscles have no role in decreasing the ROL during single leg drop landing.

**Key words:** Rate of loading · Single leg drop landing · Maximum torque of lower extremity muscles

## INTRODUCTION

The lower extremities are largely responsible for the body's ability to absorb shock during ground contact and decrease the rate of loading [1]. The rate of impact-force application, or rate of loading, is a measure of the rate of stress application to the tissues [2, 3]. High rates of loading demonstrate poor shock attenuation, indicating high stress application to the lower extremity during a short time. Factors that influence the magnitude and rate of loading include speed of movement, height, shoe type, body weight, landing-surface composition and landing strategy [2, 4-6]. Since the repetitive application of high-impact forces can lead to injury and decreased performance [1], the ability to control and adequately absorb these forces during dynamic, functional activity is the key to prevention of injury. Although it does not seen any direct relationship between magnitude and rate of impact with injurious forces, the hypothesis of relation between injurious forces with magnitude and rate of exerted impact is discussed.

Controlling the rate of loading appears significant in the high rates of repetitive loading and is associated with osteoarthritis [7]. Interestingly, the rate of loading appears to be more important than the magnitude of the load concerning joint damage [7]. In other words, similar loads or loads of even greater magnitude when applied at lower rates cause less joint damage [8] and don not promote degenerative joint disease [9]. It has been reported that microfractures, medial tibial stress syndrome, spinal injuries and other degenerative changes in joint and articular cartilage in humans to be significantly influenced by the body's ability to attenuate the associated shock from continual impacts [10, 11]. The shock experienced by the body due to landings must be attenuated by several structures and mechanisms in the body including bone, synovial fluids, cartilage, soft tissues, joint kinematics and muscular activity [12, 13]. Passively, shock attenuation is achieved by soft tissues and bone. Actively, shock attenuation is achieved through eccentric muscle action. This active mechanism is thought to be far more significant than the passive mechanism in attenuating

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shock [14]. Since muscles are thought to play a primary role in energy and shock absorption during landing, it has been hypothesized that reduced muscular torque, decreases the shock absorbing capacity of the body and subsequently can lead to an increased chance of injury [15-17].

In situation that GRF or any other external load (apart from body weight) is completely controlled by conscious muscular activity, the load is called an active load. By definition, active loads are unlikely to be harmful under normal circumstances. In everyday situations the muscles respond to changes in external loading to ensure that the body is not subjected to harmful loads [18].

However, it takes a finite time for muscles to fully respond (in terms of appropriate changes in the magnitude and direction of muscle forces) to changes in external loading. This time latency is latency period of muscles and it takes approximately 30 ms and 75 ms in adults [18]. Consequently, muscles cannot fully respond to changes in external loading that occur in less than the latency period of muscles. In these circumstances the body is forced to respond passively (by passive deformation) to the external load and, thus, this type of load is a passive load. By definition, the body is unable to control passive loads and is vulnerable to injury from high passive loads [18]. Passively, shock attenuation is achieved by soft tissues and bone. Actively, shock attenuation is achieved through eccentric muscle action. This active mechanism is thought to be far more significant than the passive mechanism in attenuating shock [19].

Recently, lower extremity muscles weakness, high Ground Reaction Forces and high ROL during normal gait have been implicated in osteoarthritis [20, 21]. Slemenda *et al* (1998) have demonstrated in cross sectional and more recently Ross *et al*. (2003) in a longitudinal analysis that quadriceps weakness is related to osteoarthritis in women [22, 23]. They reported that in women demonstrating low quadriceps strength relative to body weight, radiographic evidence of osteoarthritis developed within 2-3 years following their baseline testing. On average, the women in whom osteoarthritis developed were 15-18% weaker than those in whom it did not develop. in a study involving 32 young adults, Radin *et al*. (1991) found that individuals reporting knee pain had 37% higher rates of loading immediately after heel-strike than did those reporting no pain [20]. They theorized that individuals with high rates of loading had poor muscular strength or neuromuscular coordination, or both, which predisposed them to the development of osteoarthritis. Although high rates of loading and muscle weakness have both been linked to joint damage and pain, to our knowledge no studies have investigated whether muscle torque and ROL are associated. Our hypothesis was that there is correlation between lower extremity muscles torque and ROL during landing.

## METHODS

Thirty three healthy students with mean and standard deviation of (age 23±2 years, mass 73.24±3.16 Kg and heath of 174.51±3.40 Cm) from department of physical education and sport sciences volunteered in this study. Peak torque of quadriceps, hamstring, plantarflexors, dorsiflexors, invertors and evertors calculated by means of BIODEX dynamometer at the speed of 60 degree per second. It used verbal encouragement to exert the maximum strength from subjects. Peak torque of muscles normalized by divided to the product of weight of the subjects.

Before testing, we provided all subjects with identical instructions on the landing protocol. Subjects stood on the box in a comfortable, full weight-bearing, double-leg stance with both hands on the hips. We instructed them to drop off the box, not lower themselves from it and perform a single-leg landing on the force plate with the same leg. Upon landing, subjects were encouraged to try to maintain their balance after contact with the force plate. We allowed each subject sufficient practice trials to become comfortable with the landing procedure and to determine the preferred landing leg. The preferred landing leg was defined as the leg the subject chose to land on most frequently during the first 3 practice trials. Subjects then performed drop jumps until 5 acceptable trials were recorded. Acceptable trials were defined by the following landing criteria: (1) contact of the forefoot first, (2) maintenance of balance, (3) ability to land without hopping and (4) knee flexion less than 90°C. Subjects were not informed of the acceptable landing criteria during the test session and in no cases were more than 10 jumps required to obtain 5 acceptable trials.

The landing data are collected on force plate at a sampling rate of 200 Hz. A fast Fourier transformation analysis indicates that the raw analog signals of a single-leg stance and the jump-stabilization maneuver are below 30 Hz. Therefore, a minimum sampling rate of 60 Hz would be sufficient for collecting data. The peak GRF of the landing is a key component to calculate the ROL. A sampling rate that is too low might miss the peak force and consequently cause the ROL to be miscalculated. We selected, therefore, 200 Hz to provide a sampling rate six times greater than the raw analog-signal under study.

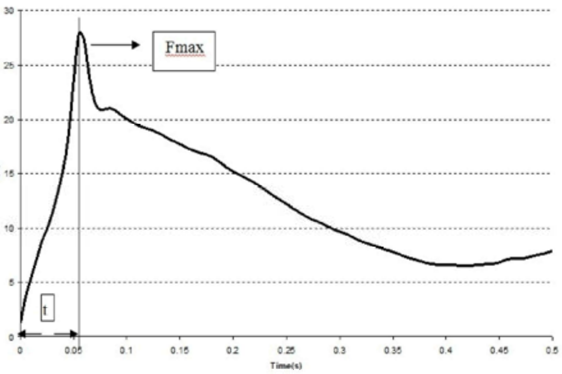

Fig. 1: Force-time diagram

Using the acquired force plate data, vertical (z direction) GRF and ROL were analyzed. We identified the first 3 acceptable trials from the 5 recorded trials and signal averaged these trials to produce a single representative trial. Trials were selected starting with the fifth trial and working backward; this ensured that all signals were accurate and representative of the landing pattern for each subject. We chose this selection method because the first trial recorded was often observably different from the remaining trials and our goal was to use trials that were most representative of the overall performance. We then used the averaged trial to measure peak vertical GRF and ROL upon landing (Figure1).We determined vertical GRF as the peak vertical force (N) recorded during landing, normalized for body weight (N) and expressed as a multiple of body weight (×BW). We measured time to peak force as the time from initial ground contact to the peak vertical force during landing. ROL was calculated as the normalized peak vertical force divided by the time to peak force.

$$ROL = \left[\frac{peakFz(N)/BW(N)}{t}\right] = \frac{BW}{ms}$$

We used Pearson coefficient method to determine the relationship between peak torques of lower extremity muscles with ROL during single leg drop landing. We used the Statistical Package for the Social Sciences (version 11.5, SPSS Inc, Chicago, IL) to analyze the data at $P \le 0.05$.

## RESULTS

Mean and standard deviation of quadriceps, hamstring, plantarflexors, dorsiflexors, invertors and evertors muscles are reported in Table 1. The results did not show any significant relationship between peak torque of Quadriceps ($r = 0.30$, $P = 0.35$), hamstring ($r = -0.21$, $P = 0.24$), plantarflexors ($r = 0.15$, $P = 0.30$), dorsiflexors ($r = 0.24$, $P = 0.25$), invertors ($r = 0.25$, $P = 0.35$) and evertors ($r = -0.46$, $P = 0.12$) muscles with ROL. Peak torques of dorsiflexors and invertors muscles are similar and the correlation coefficient between ROL and the torque of these muscles are similar as well. Although we did not see any significant relationship between peak torque of lower extremity muscles with ROL, however, regarding the table it seems that the correlation between peak torque of distal muscles with ROL is greater as compare with peak torque of proximal ones.

## DISCUSSION AND CONCLUSION

The purpose of the present research was to determine relationship between peak torques of lower extremity muscles with ROL during single leg drop landing. The results show that lower extremity muscle torque does not affect the level of ROL during single leg drop landing. Previous studies on gait [24] have shown that at 50 milliseconds after initial contact, a shock is transferred to the body due to the exchange of energy and momentum from the leg that touches the ground. The theory justifying it is that the shocks generated by GRF can be absorbed and neutralized by structures such as articular capsules, meniscus, intervertebral disc and muscles [19]. Meanwhile, the passive shock absorption takes place by soft tissues and bones and active shock absorption is done through eccentric muscle work and it is believed that the active mechanism is more important for shock absorption than the passive mechanism [19].

Regarding absorption and attenuation of GRF and ROL, it is assumed that the movements of extremities before IC can affect GRF and ROL [25]. Some people slow down the leg or stop it before it touches the ground while it seems that some others let the ground stop their leg [25]. Moreover, it is assumed that proper knee position

Table 1: Mean and S.D of ROL and Peak toque of lower extremity muscles

| | Peak torque (N.m) | | | | | | |
|---|---|---|---|---|---|---|---|
| Parameter | Quadriceps | Hamstring | Plantarflexors | Dorsiflexors | Invertors | Evertors | ROL (N/ms) |
| Mean and S.D | 2.70±0.23 | 1.75±0.25 | 0.77±0.24 | 0.43±0.15 | 0.43±0.15 | 0.40±0.12 | 519.60±172.51 |

before IC and eccentric contraction of thigh muscles at IC contribute to load dispersion and decrease stress on the joints [25]. Both the above mechanisms require a healthy muscular system for shock control.

The neuromuscular mechanism that is assumed to help prevent injuries in IC is short latency stretch reflex and immediately after IC, the body reacts to attenuate ROL. Short latency stretch reflex is generated during loading at IC by muscle spindle $I_a$-fibers and Golgi tendon organ $I_b$-afferents [26]. In any case, the time required for stretch reflex creates problems in controlling ROL. In walking, ROL and the resulting shock-wave last for about 50 milliseconds but the activation of short latency stretch reflex takes 34-42 milliseconds [25]. By the time the body has the chance to react to the step through short latency stretch reflex activity, the shock-wave would pass through calf muscles which could be a strong force attenuator [25].

The secondary mechanism for compensatory movements is related to feedback from proprioceptive receptors which were introduced by Sherrington [27]. This mechanism pertains to the awareness of the body of position (location) and movement in space. The feedback information comes from the afferent signals of the mechanoreceptors of the muscle spindles (which are distributed in the entire muscle abdomen and send the information related to muscle length and its rate of change to the nervous system), Golgi tendon organs (which are located in the muscle tendons and send information related to muscle tension and its rate of change to the nervous system), Pacinian corpuscles and Ruffini endings which are responsible for perception of the position and motion of extremities [25]. During the swing phase (a phase of the walking cycle where one leg swings forward for starting walking on the other leg), the body receives feedback from mechanoreceptors regarding the movement and with the anticipation mechanism, uses this information to maintain a controlled movement with feed forward signaling in the subsequent actions [25]. It has also been reported that if the inertia or the initial condition of extremities are no taken into consideration, the body reacts incorrectly and if the body is not aware of the movements or positions of limb segments while walking or landing from a height, it may not be able to effectively prepare for the impact and loading at IC [25].

The results of this study are limited to a static leg and cannot be generalized to jumping, cutting and other activities with change in direction. More studies are required for a thorough understanding of the role of lower extremity muscles in attenuating the impact during dynamic activities. More investigations of other factors (e.g., joint hyper mobility, leg static angle, deformations in the leg, muscle fatigue and knee and hip angles) may provide us with a better insight into the cumulative and interdependent roles of impact absorption during landing activities.

## REFERENCES


1. Hargrave, M.D., *et al*., 2003. Subtalar pronation does not influence impact forces or rate of loading during a single-leg landing. J. Athletic Training, 38(1): 18-23.
2. De Wit, B., D. De Clercq and M. Lenoir, 1995. The effect of varying midsole hardness on impact forces and foot motion during foot contact in running. J. Applied Biomechanics, 11: 395-406.
3. Cook, T.M., *et al*., 1997. Effects of Restricted Knee Flexion and Walking Speed on the Vertical Ground Reaction Force during Gait. J. Orthopaedic and Sports Physical Therapy, 25(4): 236-244.
4. Frederick, E.C. and J.L. Hagy, 1986. Factors affecting peak vertical ground reaction forces in running. International J. Sport Biomechanics, 2: 41-49.
5. Mcnitt-Gray, J.L., 1991. Kinematics and impulse characteristics of drop landings from three heights. International J. Sport Biomechanics, 7(2): 201-224.
6. Nigg, B.M. and M. Morlock, 1987. The influence of lateral heel flare of running shoes on pronation and impact forces. Medicine and Science in Sports and Exercise, 19(3): 294-302.
7. Mikesky, A.E., A. Meyer and K.L. Thompson, 2000. Relationship between quadriceps strength and rate of loading during gait in women. J. Orthopaedic Research, 18(2): 171-175.
8. Radin, E., *et al*., 1985. Mechanical factors influencing cartilage damage. Osteoarthritis: Current Clinical and Fundamental Problems., pp: 90-99.
9. Michel, B., *et al*., 1992. Osteophytosis of the knee: association with changes in weight-bearing exercise. Clinical rheumatology, 11(2): 235-238.
10. Light, L., G. McLellan and L. Klenerman, 1980. Skeletal transients on heel strike in normal walking with different footwear. J. Biomechanics, 13(6): 477-480.
11. McMahon, T.A., G. Valiant and E.C. Frederick, 1987. *Groucho running.* J. Applied Physiol., 62(6): 2326-37.
12. Lafortune, M.A., M.J. Lake and E.M. Hennig, 1996. Differential shock transmission response of the human body to impact severity and lower limb posture. J. Biomechanics, 29(12): 1531-7.

13. Nyland, J.A., *et al*., 1994. Relationship of fatigued run and rapid stop to ground reaction forces, lower extremity kinematics and muscle activation. J. Orthopedic Sports and Physical Therapy, 20(3): 132-7.
14. Mizrahi, J. and Z. Susak, 1982. Analysis of parameters affecting impact force attenuation during landing in human vertical free fall. Eng Med., 11(3): 141-7.
15. Radin, E.L., 1986. Role of muscles in protecting athletes from injury. Acta. Med. Scand Suppl, 711: 143-7.
16. Verbitsky, O., *et al.*, 1998. Shock transmission and fatigue in human running. J. Applied Biomechanics, 14: 300-311.
17. Voloshin, A.S., *et al*., 1998. Dynamic loading on the human musculoskeletal system--effect of fatigue. Clinical Biomechanics, 13(7): 515-520.
18. Watkins, J., 2009. Structure and function of the musculoskeletal system. Human Kinetics.
19. Coventry, E., *et al*., 2006. The effect of lower extremity fatigue on shock attenuation during single-leg landing. Clinical Biomechanics, 21(10): 1090-1097.
20. Radin, E.L., *et al.*, 1991. Relationship between lower limb dynamics and knee joint pain. J. Orthopaedic Res., 9(3): 398-405.
21. Slemenda, C., *et al.*, 1997. Quadriceps weakness and osteoarthritis of the knee. Annals of Internal Medicine, 127(2): 97-104.
22. Slemenda, C., *et al*., 1998. Reduced quadriceps strength relative to body weight: a risk factor for knee osteoarthritis in women? Arthritis and Rheumatism, 41(11): 1951-1959.
23. Ross, S. and K. Guskiewicz, 2003. Time to stabilization: a method for analyzing dynamic postural stability. Athletic Therapy Today, 8(3): 37-39.
24. Wakeling, J.M., A.M. Liphardt and B.M. Nigg, 2003. Muscle activity reduces soft-tissue resonance at heel-strike during walking. J. Biomechanics, 36(12): 1761-1769.
25. Riskowski, J., *et al*., 2005. Proprioception, gait kinematics and rate of loading during walking: Are they related? J. Musculoskeletal and Neuronal Interactions, 5(4): 379-387.
26. Yakovenko, S., V. Gritsenko and A. Prochazka, 2004. Contribution of stretch reflexes to locomotor control: a modeling study. Biological Cybernetics, 90(2): 146-155.
27. Sherrington, C., 1947. The integrative action of the nervous system. CUP Archive.