MILITARY MEDICINE, 174, 8:832, 2009

Does Foot Pitch at Ground Contact Affect Parachute Landing Technique?

John W. Whitting, BSc(Hon); Julie R. Steele, PhD*; Mark Jaffrey, PhD†; Bridget J. Munro, PhD**

ABSTRACT The Australian Defence Force Parachute Training School instructs trainees to make initial ground contact using a flat foot whereas United States paratroopers are taught to contact the ground with the ball of the foot first. This study aimed to determine whether differences in foot pitch affected parachute landing technique. Kinematic, ground reaction force and electromyographic data were analyzed for 28 parachutists who performed parachute landings (vertical descent velocity = 3.4 $m \cdot s^{-1}$) from a monorail apparatus. Independent *t*-tests were used to determine significant ($p < 0.05$) differences between variables characterizing foot pitch. Subjects who landed flat-footed displayed less knee and ankle flexion, sustained higher peak ground reaction forces, and took less time to reach peak force than those who landed on the balls of their feet. Although forefoot landings lowered ground reaction forces compared to landing flat-footed, further research is required to confirm whether this is a safer parachute landing strategy.

INTRODUCTION

Military static-line parachuting is designed to deliver large numbers of armed personnel to a battlefield, quickly, safely, and ready to engage the enemy.[1] Upon landing from a static-line parachute descent, paratroopers are forced to absorb extremely high impact loads, thereby making this form of aerial descent among the most hazardous,[2,3] with a well-documented injury risk currently averaging 6 injuries per 1,000 jumps.[1] Because of the excessive impact forces and the rapid rate of loading sustained when a paratrooper lands, the lower limbs and vertebral column are at most risk of injury.[1,4] In fact, lower limb injuries have been reported to be as high as 80% of the total parachuting injuries incurred[5] and poor landing technique has been identified as the single largest cause of these injuries.[5,6]

The 5-point parachute landing fall (PLF) technique is a common landing method taught to paratroopers worldwide in an attempt to minimize the risk of injury.[1,6] During the final stage of an aerial descent, a paratrooper assumes a prelanding posture involving slight flexion of all lower limb joints to allow for optimal absorption of the initial ground reaction forces (GRF).[7] Subtle between-nation technique differences exist, however, with regard to foot pitch at initial ground contact (IC). The Australian Defence Force Parachute Training School (PTS) instructs trainees to land with a flat foot, whereas United States (U.S.) paratroopers are taught to present the ball of the foot to the ground as the first point of contact.[1,8] Following IC by the feet, the paratrooper absorbs the initial impact of the landing by eccentrically dorsiflexing the ankle and flexing the knee and hip, before performing a standardized PLF roll using multiple points of body–ground contact. The PLF roll minimizes the need to dissipate all of the GRF via the lower limbs by enabling paratroopers to distribute the load over as much of the body as possible.[6]

Vertical descent velocities achieved by trainees during ground training at the Australian Defence Force PTS range between 2.1 and 3.4 $m \cdot s^{-1}$, while aerial parachute descents onto the drop zone result in vertical descent velocities of between 4.6 and 6.7 $m \cdot s^{-1}$. A study involving U.S. paratroopers has shown that simulated parachute landings with vertical descent velocities within this range can result in the generation of GRF in excess of 17 times body weight.[9] It is widely accepted that lower limb loading increases with increases in both landing velocity and leg stiffness.[10,11] In static-line parachuting, however, descent velocity is not easily reduced, making efficient lower limb mechanics crucial in terms of moderating the high-impact forces.

The lower limb is commonly regarded as a spring, used to rapidly and eccentrically absorb external loading[12] via a complex multijoint solution.[13] A low joint range of motion, particularly in the ankle, knee, and hip during impact absorption, is generally associated with a stiff landing.[10] Butler et al.[10] stated that an optimal level of leg stiffness was required to avoid injury during landing activities. These researchers suggested that too much leg stiffness may be associated with bony injuries, because of the rapid rate of loading and higher forces, and too little leg stiffness may be associated with a larger excursion of the joints, leading to soft tissue injuries.[10] Research has also shown that by adjusting lower limb mechanics in a drop landing, leg stiffness may be altered to compensate for changes in landing surface to maintain an optimal level of landing stiffness.[14] It should also be noted that ankle stiffness has been shown to moderate overall lower limb stiffness during single-legged hopping,[15] whereby increased ankle stiffness was correlated with a proportional increase in leg stiffness and GRF.

Research has shown that paratroopers display different biomechanical strategies when performing simulated PLF

*Biomechanics Research Laboratory, School of Health Sciences, University of Wollongong, Northfields Avenue, Wollongong, NSW, 2522, Australia.

†Human Protection and Performance Division, Defence Science and Technology Organisation, Department of Defence, 506 Lorimer Street, Fishermans Bend, VIC, 3207, Australia.

This manuscript was received for review in September 2008. The revised manuscript was accepted for publication in May 2009.

landings using different rates of vertical descent velocity,[9,16] different drop heights,[13,17,18] and with different foot pitch.[10,19] However, no researchers have systematically investigated the effects of variations in foot pitch at IC on PLF technique under realistic ground training conditions of vertical and horizontal descent velocities. Therefore, the purpose of the present study was to examine the effects of variations in foot pitch at initial impact on the PLF technique. It was hypothesized that subjects who landed with a flat-footed pitch at initial impact would generate higher GRF, with less time taken to reach the peak GRF. It was also hypothesized that flat-footed landings would be associated with less ankle range of motion, albeit greater knee range of motion as subjects attempted to moderate overall lower limb stiffness during load absorption.

MATERIALS AND METHODS

Twenty-eight Basic Parachute Course trained personnel (27 male, 1 female; mean age = 30 ± 7 years; height = 177.0 ± 6.8 cm; mass = 84.2 ± 13.4 kg) were recruited from the Australian Defence Force PTS, New South Wales, Australia, as volunteer subjects for this study. Only paratroopers who were cleared to jump by military medical personnel were included in the study. Ethical clearance for the study was obtained from the University of Wollongong Human Research Ethics Committee and the Australian Defence Human Research Ethics Committee.

The Experimental Task

After providing informed consent, the subjects suspended themselves from a free-wheeling trolley of a custom-designed monorail system[16] (see Fig. 1). The trolley handle was telescopic, allowing for easy adjustment of the height of the paratroopers' feet from the ground while they assumed the typical parachute landing posture before release of the handle. This handle adjustment enabled the experimenters to standardize drop height at 74 cm; thereby standardizing descent velocity on the basis of calculations made using standard equations of motion. It should be noted that a slightly lower vertical descent velocity was achieved as the subjects tended to lower their release foot height by slight knee extension and grip adjustment during the trials. The final vertical descent velocity was measured using two digital video cameras that were synchronized with a ground reaction force platform (see section on Parachute Landing Fall Technique) and as such, each subject performed a minimum of five successful PLF trials at a vertical descent velocity of 3.3 ± 0.2 $m \cdot s^{-1}$, with a constant horizontal drift velocity of 2.3 ± 0.01 $m \cdot s^{-1}$.

The subjects were not given any feedback on prelanding foot pitch and self-selected either a flat-footed (FF) or ball-of-foot first (BF) orientation in preparation for initial impact absorption. Throughout the trials the subjects wore their standard army boots and safety helmet but wore Lycra shorts and a firmly fitted T-shirt to enhance observation of body alignment during the experimental task.

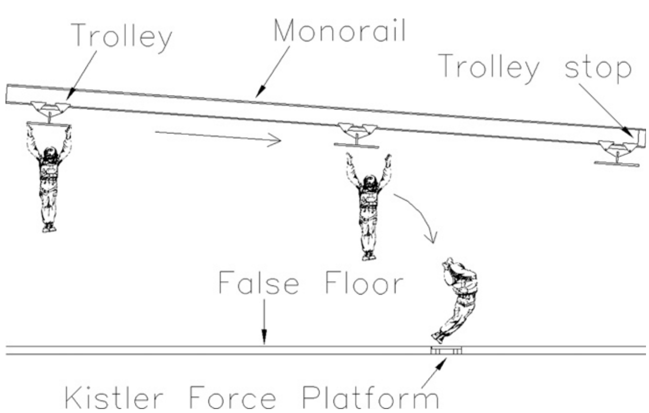

FIGURE 1. Overhead monorail used to simulate realistic ground training and drop zone parachute landing descent rates. Subjects were suspended below a free-wheeling trolley, which traveled along the monorail at a decline of approximately 4°. Markers on the landing strip were used as visual cues to prompt subjects to release their hands from the trolley to land successfully on the force platform. Trolley handles were adjustable so that foot height was standardized for all subjects to achieve the correct descent velocity.

Parachute Landing Fall Technique

The PLF technique of each subject was captured (25 Hz; shutter speed 1/4,000 s) using two digital video cameras (DCR-TRV50E, Sony Corporation, Japan). One camera was positioned to capture handle release while the other camera was positioned to record motion of the lead lower limb in the sagittal plane. The cameras were synchronized with the force and electromyographic (EMG) data using a light emitting diode, positioned in the camera field of view, which sent a synchronization pulse through the customized LabVIEW (National Instruments Corporation, Austin, TX, Version 7.1) software. Acknowledging the limited film rate capacity of the available video equipment, the video images were primarily used to qualitatively characterize the PLF technique used by each subject during each descent velocity condition. This was completed by visually comparing the foot pitch displayed at IC and the number of body–ground contact points achieved during execution of the PLF roll against relevant aspects of the Parachute Jump Instructor Checklist used during Basic Parachute Courses at the Australian Defence Force PTS.[20]

The video images were also used to assess each subject's two-dimensional lower limb alignment in the sagittal plane immediately before IC and at the end of the initial impact absorption (IA) phase as a preliminary estimate on how the subjects used their lower limbs to absorb the impact forces. To facilitate later analysis of the video images, the joint centers of the leading hip (greater trochanter), knee (immediately superior to the fibular head), and ankle (lateral malleolus) of each subject were highlighted with adhesive markers. Once the video images were captured, these markers were used to quantify absolute knee and ankle angles (°) during the IA phase (°) using Silicon COACH PRO (Silicon Coach Ltd., Dunedin, New Zealand, Version 6) analysis software.

Ground Reaction Force Data

The GRF generated by each subject at landing were quantified using a Kistler force platform (Type 9281B, Kistler Instrumente AG Winterthur, Switzerland; 600 mm × 400 mm) isolation mounted to the concrete floor and surrounded by a timber landing platform. Although kinematic and force data are affected by landing surface,[14,18] both the force platform and surrounding surface were covered with 20-mm thick EVA rubber as a safety precaution.

Force data were sampled (1,000 Hz) for each successful PLF trial, whereby the subject landed both feet within the confines of the force platform. All force data were channeled through a National Instruments (DAQpad 6015/1016) break-out box and collected using customized LabVIEW software. The summed GRF data were processed to calculate the time of IC, the peak vertical ground reaction force (F_Y), the peak resultant ground reaction force (F_R), and the time to the F_R (ms). The force values were reported both in Newtons (N) as well as normalized for body weight (BW).

Muscle Activation Data

Due to their role in absorbing landing forces,[7,21,22] muscle activation patterns were recorded from six superficial lower limb muscles of the lead lower limb: medial gastrocnemius (MG), tibialis anterior (TA), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), and semitendinosus (ST). After standard preparation to reduce electrical impedance of the skin below 6 k Ohms (CardioMetrics Artifact Eliminator, Australia), Blue Sensor (SP-00-S) bipolar silver–silver chloride surface electrodes (20 mm interelectrode spacing) were fixed over the relevant muscle bellies and aligned parallel to the underlying muscle fiber direction while a reference electrode was placed on the medial femoral epicondyle. The Telemyo 900 system (Noraxon) housed in a pack firmly secured to the chest of each subject sampled the EMG data at 1,000 Hz (bandwidth 16–500 Hz).

After removing signal offset, the EMG signals were filtered using a fourth-order zero-phase-shift Butterworth filter (high-pass f_c = 15 Hz)[23] to eliminate any movement artifact. To quantify temporal characteristics of the muscle bursts, the filtered EMG data were full-wave rectified and then filtered using a fourth-order zero-phase-shift Butterworth low-pass filter (f_c = 20 Hz) to create linear envelopes, resulting in a smoothed representation of the EMG signals. The linear envelopes closely resembled the shape of the muscle tension curves while retaining the signals' critical temporal characteristics.[24] Individual linear envelopes representing each muscle burst were then screened using a threshold of 8% of the maximum amplitude to determine muscle onset and offset relative to IC and muscle burst duration. The filtered EMG signal of each individual muscle was then visually inspected to confirm the validity of the calculated results.

Statistical Analysis

The subjects were divided into two foot-pitch groups according to whether they consistently landed (four or more of the five successful landings) on either the balls of their feet (BF; n = 19) or flat-footed (FF; n = 9). A Kolmogorov–Smirnov test (with Lilliefor's correction) was used to test data for normality before analysis. An independent samples t-test design was used to determine whether there were any significant ($p < 0.05$) differences between the kinematic, GRF, or EMG variables analyzed for each group. Although multiple t-tests were conducted, an α level adjustment was deemed unnecessary given the exploratory nature of the study. All statistical analyses were conducted using SPSS software (SPSS 11.5 for Windows; SPSS Inc., Chicago, IL).

RESULTS

PLF Technique

During the 134 trials analyzed for the study, 19 subjects landed with a BF orientation, while 9 subjects landed with a FF pitch on at least 4 occasions each, with BF landings accounting for 69% of the total landings. Interestingly, the tendency for these Australian paratroopers to land BF first is in direct contrast with the FF technique taught during Australian Defence Force Basic Parachute Courses.[8]

Descriptive data pertaining to the kinematic variables characterizing the PLF technique displayed by the subjects are presented in Table I. A significant difference was found between

TABLE I. Means (± SD), t-Values, and α Levels Derived for the Effects of Foot Pitch on the Kinematic Data Displayed by the Subjects

	Foot Pitch		Statistics	
Variable[a]	Flat-Footed (n = 9)	Ball of Foot (n = 19)	t-Value	p-Value
Knee Angle at IC (°)[b]	154 ± 4	159 ± 4	−2.544	0.022
Knee Angle at End of IA Phase (°)	106 ± 8	103 ± 7	0.730	0.477
Knee ROM (°)[b]	49 ± 6	55 ± 6	−2.705	0.016
Ankle Angle at IC (°)[b]	101 ± 4	126 ± 8	−11.119	<0.001
Ankle Angle at End of IA Phase (°)[b]	93 ± 4	100 ± 10	−2.666	0.013
Ankle ROM (°)[b]	8 ± 5	26 ± 12	−5.594	<0.001
Time for IA Phase (ms)[b]	68 ± 13	88 ± 11	−4.143	0.001
Time for 5 Points of PLF (ms)	711 ± 122	761 ± 127	−0.988	0.337

[a] IC, initial contact; IA, initial impact absorption phase (from IC to the point of maximum knee flexion before the PLF roll); ROM, range of motion. [b] Indicates a significant difference between the foot pitch conditions.

the BF and FF groups, for nearly all kinematic variables displayed during the IA phase. That is, subjects in the BF group contacted the ground with significantly greater knee extension and ankle plantar flexion and displayed a larger range of knee and ankle motion during impact absorption. However, interestingly, there was no difference seen in the knee angle at the completion of the IA phase. Furthermore, the BF group took significantly longer during the IA phase, although no difference was seen between the groups in the time to complete the overall PLF roll.

Ground Reaction Forces

A significant between-group difference was found for all GRF variables analyzed for the subjects when performing the PLF (see Table II). That is, subjects in the FF group displayed significantly higher vertical GRF in both absolute terms and when normalized for BW. Furthermore, the FF group took significantly less time to reach the F_R, indicating a more rapid rate of loading than the BF group.

Muscle Activation Patterns

Descriptive data pertaining to the onset and duration of each of the six lower limb muscles monitored for both subject groups during the PLF are summarized in Table III. There were no significant between-group differences in the timing of any of the six muscles, except for MG onset, which was significantly earlier, and ST offset, which was significantly later for the BF group relative to their FF counterparts. However, these results must be interpreted with caution, because of a consistently large impact artifact in the EMG data that reduced the number of trials and subjects for which EMG data were suitable for analysis.

DISCUSSION

Kinematics

Although all subjects had been trained at the Australian Defence Force PTS to use the FF landing technique, 69% landed with a BF foot pitch. This technique modification added

TABLE II. Means (± SD), *t*-Values and α Levels Derived for the Effects of Foot Pitch on the Ground Reaction Force Variables Displayed by the Subjects

	Foot Pitch		Statistics	
Variable[a]	Flat-footed (n = 9)	Ball of foot (n = 19)	*t*-Value	*p*-Value
Peak F_Y (N)[b]	9087 ± 1467	7141 ± 1440	3.296	0.005
Peak F_R (N)[b]	9422 ± 1483	7467 ± 1479	3.261	0.005
Peak F_Y (BW)[b]	10.8 ± 0.8	8.4 ± 1.4	5.587	<0.001
Peak F_R (BW)[b]	11.2 ± 0.8	8.8 ± 1.5	5.489	<0.001
IC to Peak F_R (ms)[b]	20.1 ± 2.1	37.4 ± 5.9	−11.373	<0.001

[a] F_Y, vertical ground reaction force; F_R, resultant ground reaction force; IC, initial ground contact. [b] Indicates a significant difference between the foot pitch conditions.

TABLE III. Means (± SD), *t*-Values and α Levels Derived for the Effects of Foot Pitch on the Onset, Offset and Burst Duration (ms) for the Six Lower Limb Muscles Displayed by the Subjects

	Foot Pitch				Statistics	
Variable	Flat-Footed	n[a]	Ball of Foot	n[a]	*t*-Value	*p*-Value
TA Onset	−98 ± 98	9	−118 ± 115	19	0.464	0.648
TA Offset	783 ± 217	9	788 ± 343	18	−0.048	0.962
TA Duration	879 ± 217	9	908 ± 374	18	−0.252	0.803
MG Onset[b]	−117 ± 67	7	−222 ± 87	17	3.181	0.006
MG Offset	366 ± 163	3	362 ± 172	7	0.036	0.973
MG Duration	508 ± 162	3	579 ± 237	7	−0.55	0.603
RF Onset	−116 ± 36	9	−139 ± 121	17	0.745	0.465
RF Offset	482 ± 324	9	417 ± 271	17	0.509	0.618
RF Duration	597 ± 307	9	557 ± 186	17	0.361	0.725
VM Onset	−125 ± 39	9	−123 ± 41	17	−0.109	0.915
VM Offset	345 ± 142	9	394 ± 166	17	−0.773	0.449
VM Duration	470 ± 162	9	515 ± 198	17	−0.624	0.54
BF Onset	−86 ± 51	9	−103 ± 35	14	0.877	0.396
BF Offset	406 ± 156	9	395 ± 97	12	0.191	0.852
BF Duration	490 ± 194	9	500 ± 95	12	−0.143	0.889
ST Onset	−137 ± 54	6	−105 ± 38	10	−0.125	0.247
ST Offset[b]	432 ± 85	5	556 ± 87	8	−2.542	0.032
ST Duration	574 ± 117	5	668 ± 113	8	−1.427	0.19

[a] Because of excessive motion artifact in the EMG signal at impact, it was not possible to analyze EMG variables for all subjects. Therefore, caution is required when interpreting variables with low subject numbers. [b] Indicates a significant difference between the foot pitch conditions.

an extra segment to the IA phase and it is postulated that this would have greatly influenced absorption of the GRF.[16] Self and Paine[19] described initial ankle angles of 127 to 130° being displayed in drop landings where subjects were instructed to use their toes to absorb some of the landing impact. The average ankle angle at IC for the BF group was 126° (see Table I), indicating similarities in ankle kinematics at IC between BF landings and drop landings during the IA phase of the PLF. The FF group averaged an ankle angle at IC of only 101°, with a range of only 8° during the IA phase. This contrasts dramatically with a much larger range of ankle angular displacement for the BF group (26°), albeit occurring with virtually no dorsiflexion beyond the neutral flat foot, as was hypothesized.

Interestingly, our hypothesis that the BF group would compensate for a greater ankle range of motion by minimizing knee range of motion, was not supported. In fact, relative to the FF subjects, the BF group displayed significantly more knee extension at IC and employed a larger range of knee motion during the IA phase, albeit displaying a similar final knee angle to the FF group at the end of the IA phase. This is consistent with previous research that suggests landing requires a complete multijoint solution,[14] that kinematic control in a movement task depends on the orientation of body segments relative to external loading[25] and may actually be moderated by ankle kinematics.[15] That is, each foot-pitch group used an entirely different mechanical strategy to absorb the initial impact forces and, although no actual measures of leg stiffness were made, these kinematic results suggest that the BF group exhibited a much softer landing strategy.

There was no statistical difference between the BF or FF groups in terms of the time taken to make the standard 5 points of PLF body–ground contact (see Table I), irrespective of the apparent disparity in landing stiffness during the IA phase. Furthermore, subjects in each group took, on average, over 700 ms to complete the initial 5 points of contact, which is longer than the 400 ms suggested to be indicative of a good PLF roll, absent of potentially injurious technique faults.[7] If overall forces imposed on the body during a PLF roll are to be moderated by the time taken to complete the roll, then the present study supports the notion that paratroopers may be able to moderate overall landing impact by adjusting the initial impact absorption strategy. However, no GRFs were sampled beyond the IA phase in the present study, so actual forces imposed on the body during the rolling phase remain unknown.

Ground Reaction Forces

The mean GRF values of between 8.4 and 11.2 BW (see Table II) generated during the present study are relatively high when compared with GRF data for other landing movements reported at similar descent velocities, which show GRF values of up to approximately 6 BW.[13,22,25,26] Values commensurate with, and also in excess of, the GRF found in this study have been reported in the literature, although these studies involved higher vertical descent velocities than were tested in the present study.[9,13,27,28] Despite between-study differences in GRF generated at landing, there is universal agreement that vertical GRF increases in direct response to increases in drop height and descent velocity.[11,18]

Results from the present study, however, show that factors other than descent velocity also affected GRF. The BF group recorded an average F_Y value of 8.4 BW compared to 10.8 BW for the FF group, which supports our original hypothesis that flat-footed landings would generate higher GRF (see Table II). As stated above, differences in lower limb mechanics between the two experimental groups appear responsible for altering the magnitude of the impact force. Humans display an ability to predict impact forces[29] and employ complex mechanical multijoint solutions,[14] such as increases in joint range of motion, to allow effective attenuation of loads experienced at IC.[11] In the present study, the degree of knee joint motion displayed by subjects seems to have been moderated by the ankle joint,[10,15] thereby leading to lower GRF in the BF group. Perhaps the lower limb mechanics displayed by the BF subjects during the IA phase were part of an overall strategy to moderate forces imposed on other parts of the body during subsequent phases of the PLF roll by extending the time of the IA phase.

The difference in landing mechanics displayed by each experimental group in the present study is further elucidated by the time taken to reach the peak F_R. The BF group took almost twice as long to reach the peak F_R as the FF group (see Table II), supporting our original hypothesis and demonstrating that FF landings led to not only higher GRF, but also a significantly faster rate of loading. This finding supports other research, which suggests an inverse relationship exists between the magnitude of the peak GRF and the time to reach the peak forces.[18] Studies have also shown that landings that are too stiff, with high GRF and rates of loading, are associated with higher rates of catastrophic bony injuries.[10] The PLF requires a strategy that accounts for impact forces spread throughout an extended movement beyond the initial impact encountered by the lower limbs. By adding the extra segment of a plantar flexed foot, the BF group was able to increase impulse time during the IA phase, thereby reducing the absolute load in this phase of the landing.

Muscle Activation Patterns

Because of excessive movement artifact at or around the time of IC, it was only possible to analyze temporal aspects of muscle activity during the PLF. However, the muscle onset results in the present study were consistent with previous research for landings performed with similar descent velocities.[22,29,30] Furthermore, with only 2 of the 18 EMG variables showing a significant difference between the BF and FF groups, it appears that a reasonably consistent neuromuscular strategy exists in terms of prelanding preparation and initial impact absorption. The consistent recruitment of the antigravity extensors earlier than the antagonist stabilizers (see Table III), irrespective of prelanding foot pitch, load absorption kinematics, or resultant

GRF, may indicate the need to resist joint flexion during the initial impact absorption phase.[21] This supports the notion that the motor control system behaves in a predictive manner to set the level of limb compliance deemed appropriate for the ensuing impact force[29] and that the underlying mechanism might be a preprogrammed, feed-forward mode of control.[22] The earlier onset of MG in the BF group may be a predictive mechanism to prepare for eccentric absorption of force through a larger range of ankle motion. Similarly, it is postulated that a later ST offset in the BF group could be the result of the extended time for the IA phase as the knee is stabilized for a longer duration and over a larger range of motion.

CONCLUSIONS

The IA phase of the PLF technique was significantly affected by prelanding foot pitch. Despite being trained to hold a neutral flat-footed posture at ground contact in preparation to perform a PLF, many of the subjects plantar flexed their feet to contact the ground with the balls of the feet. As such, the BF group extended the knees and plantar flexed the feet more at IC than the FF group and used significantly greater knee and ankle range of motion during the IA phase, leading to a lower peak F_R and a slower rate of loading during the IA phase. It is postulated that the technique modifications used by the BF group were protective adaptations to reduce excessive lower limb loading during IC, even though the overall impulse time for the entire PLF movement was similar for both groups. Although the actual magnitude or rate of loading which can be optimally or safely absorbed by the lower limbs in a complex landing movement such as the PLF is not yet understood, because of the excessive GRF and loading rates encountered during FF landings, it is speculated that this foot orientation may lead to a higher risk of lower limb injury than a BF landing. Therefore, it is recommended that future research should address whether modification to the PLF technique being taught to paratroopers at the Australian Defence Force PTS is warranted, particularly with respect to foot pitch at ground contact.

ACKNOWLEDGMENTS

This work was supported by the Australian Government, Department of Defence.

REFERENCES

1. Knapik JJ, Craig SC, Hauret KG, Jones BH: Risk factors for injuries during military parachuting. Aviat Space Environ Med 2003; 74: 768–74.
2. Ellitsgaard N: Parachuting injuries: a study of 110,000 sports jumps. Br J Sports Med 1987; 21: 13–7.
3. Bricknell MC, Amoroso PJ, Yore MM: What is the risk associated with being a qualified military parachutist? Occup Med 1999; 49: 139–45.
4. Ellitsgaard N, Warburg F: Movements causing ankle fractures in parachuting. Br J Sports Med 1989; 23: 27–9.
5. Ekeland A: Injuries in military parachuting: a prospective study of 4499 jumps. Injury 1997; 28: 219–22.
6. Bricknell MC, Craig SC: Military parachuting injuries: a literature review. Occup Med 1999; 49: 17–26.
7. Henderson JM, Hunter SC, Berry WJ: The biomechanics of the knee during the parachute landing fall. Milit Med 1993; 158: 810–6.
8. Manual of Land Warfare 3.2.6: Parachute Training: Headquarters Training Command, Australian Army, 1993.
9. Crowell HP, III, Treadwell TA, Faughn JA, Leiter KL, Woodward AA, Yates CE: Lower Extremity Assistance for Parachutists (LEAP) Program: Quantification of the Biomechanics of the Parachute Landing Fall and Implications for a Device to Prevent Injuries. Aberdeen Proving Ground: U.S. Army Research Laboratory; November 1995, ARL-TR-926.
10. Butler RJ, Crowell HP 3rd, Davis IM: Lower extremity stiffness: implications for performance and injury. Clin Biomech 2003; 18: 511–7.
11. Zhang SN, Bates BT, Dufek JS: Contributions of lower extremity joints to energy dissipation during landings. Med Sci Sports Exerc 2000; 32: 812–9.
12. Devita P, Skelly WA: Effect of landing stiffness on joint kinetics and energetics in the lower extremity. Med Sci Sports Exerc 1992; 24: 108–15.
13. McNitt-Gray JL: Kinetics of the lower extremities during drop landings from three heights. J Biomech 1993; 26: 1037–46.
14. McNitt-Gray JL, Yokoi T, Millward C: Landing strategy adjustments made by female gymnasts in response to drop height and mat composition. J Appl Biomech 1993; 9: 173–90.
15. Farley CT, Morgenroth DC: Leg stiffness primarily depends on ankle stiffness during human hopping. J Biomech 1999; 32: 267–73.
16. Whitting JW, Steele JR, Jaffrey MA, Munro BJ: Parachute landing fall characteristics at three realistic vertical descent velocities. Aviat Space Environ Med 2007; 78: 1135–42.
17. Hoffman JR, Liebermann D, Gusis A: Relationship of leg strength and power to ground reaction forces in both experienced and novice jump trained personnel. Aviat Space Environ Med 1997; 68: 710–4.
18. Seegmiller JG, McCaw ST: Ground reaction forces among gymnasts and recreational athletes in drop landings. J Athl Train 2003; 38: 311–4.
19. Self BP, Paine D: Ankle biomechanics during four landing techniques. Med Sci Sports Exerc 2001; 33: 1338–44.
20. Jaffrey M, Steele JR: Landings During 2004 Basic Parachute Courses: Common Technique Faults and Injuries: Australian Government, Department of Defence, Defence Science and Technology Organisation, Melbourne. Technical Report, 2007. DSTO-TR-2055 AR-014-028.
21. Caster BL, Bates BT: The assessment of mechanical and neuromuscular response strategies during landing. Med Sci Sports Exerc 1995; 27: 736–44.
22. Liebermann DG, Hoffman JR: Timing of preparatory landing responses as a function of availability of optic flow information. J Electromyogr Kinesiol 2005; 15: 120–130.
23. Winter DA: Biomechanics and Motor Control of Human Movement, Ed 2, Toronto, John Wiley & Sons, 1990.
24. Cowling EJ, Steele JR: Is lower limb muscle synchrony during landing affected by gender? Implications for variations in ACL injury rates. J Electromyogr Kinesiol 2001; 11: 263–8.
25. McNitt-Gray JL, Hester DM, Mathiyakom W, Munkasy BA: Mechanical demand and multijoint control during landing depend on orientation of the body segments relative to the reaction force. J Biomech 2001; 34: 1471–82.
26. Dufek JS, Bates BT: The evaluation and prediction of impact forces during landings. Med Sci Sports Exerc 1990; 22: 370–7.
27. Oggero E, Pagnacco G, Morr DR, Barnes SZ, Berme N: The mechanics of drop landing on a flat surface–a preliminary study. Biomed Sci Instrum 1997; 33: 53–8.
28. Santello M, McDonagh MJ, Challis JH: Visual and non-visual control of landing movements in humans. J Physiol 2001; 537: 313–27.
29. Santello M, McDonagh MJ: The control of timing and amplitude of EMG activity in landing movements in humans. Exp Physiol 1998; 83: 857–74.
30. Arampatzis A, Morey-Klapsing G, Bruggemann GP: The effect of falling height on muscle activity and foot motion during landings. J Electromyogr Kinesiol 2003; 13: 533–44.