DESIGN AND EVALUATION OF PROTECTIVE DEVICES FOR INJURY PREVENTION DURING PARATROOPER LANDING

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ABSTRACT

An integrated approach including laboratory tests and numerical simulation was applied to investigate the mechanical behavior of paratroopers during the parachute landing fall (PLF) in order to reduce related injuries. This paper presents the development and validation of the biomechanical finite element (FE) model used to simulate paratrooper landings and analyze related injuries. Latest applications of the model include investigation of tailored designs of protective devices such as ankle braces on lower extremity injuries. Biofidelity improvements on the model such as proper joint stiffness provided by test validation are also detailed. A complete paratrooper model is simulated for various landing scenarios for injury analysis, which provides a safe, inexpensive and easy-to-use tool to understand the dynamics of paratrooper landing compared to physical testing. The model was also used to evaluate ankle brace design, terrain slope and landing speed through parametric studies.

INTRODUCTION

Paratroopers have long been among those in the Army at the highest risk of serious injuries. They are more vulnerable to injury than their civil counterparts because of heavy equipment they carry on the job. The risk of injury is further increased due to the frequency of military operations in bad weather and darkness, and in dangerous terrain. Most of the parachute landing injuries involves the lower extremity of the human body. Studies reported that this accounts for an average of more than 60 percent of parachuting injuries. Among lower extremity injuries, ankle injuries represent the predominant injury in both civilian and military parachutists, ranging from 30 to 60 percent. Other lower extremity injuries include knee sprains of various ligaments and muscles as well as ankle, foot and leg fractures. These injuries are due to excessive impact forces and moments, mostly from poor landings. Pelvic and spinal fractures and head injuries are due to poor backward landing. Most injuries occur during landing and can be avoided with proper landing techniques.

Acute and disabling injuries are a great concern for military operation, since soldiers who cannot get off a drop zone by themselves are ineffective and vulnerable to enemy fire or capture. Injured soldiers may also require assistance of other soldiers to get off the drop zone. This can potentially reduce the effectiveness of an airborne assault force. With the desire for lower altitude and higher speed parachute drops in the future, parachuting injuries will become an even more serious threat to efficient military operations unless effective countermeasures are considered and implemented. Soldiers parachuting from lower attitudes and at higher speeds are likely to have higher landing velocities, and they will probably have less control over their orientation with respect to the ground during landing.

Various protective devices including ankle and knee braces, energy absorbing soles are evaluated to reduce paratrooper landing injuries. The braces are useful to prevent excessive rotation of ankle and knee joints, and the energy absorbing soles to reduce impact force. The tested results in this report illustrate the effectiveness of several protective devices. However, it is important to note that due to practical limitations of performance evaluation of foot wears based on human testing, it is difficult to achieve optimal designs of protective foot wears. In addition, many scenarios including landing on rough terrain and new landing techniques involve possible injuries for participants.

A complementary and effective approach to testing and evaluating the parachute landing injuries and protective foot wears is to model paratrooper landings using modern simulation techniques. With the increasingly rapid development of high power computing systems, computer modeling is becoming a more realistic and reliable method in biomechanics research. With a validated

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biomechanical model, the dynamic responses of paratroopers under various landing scenarios can be simulated and injury risks evaluated. New devices can be evaluated even before their prototypes are developed, reducing lead-time for product development.

This project is a continuation of the previous investigation on this subject, which presented the development of a finite element model of paratrooper landing based on GEBOD data. This paper describes the addition of protective foot wears, and parametric studies that were conducted on the various landing conditions and landing speeds.

LABORATORY TESTING

Test Setup

PLF tests were conducted in the Biomechanics Laboratory in the Department of Exercise Science at the University of Massachusetts – Amherst. This paper describes the third test conducted. The experimental setup was very similar to the previous two tests [1], which involved collection of ground reaction forces, lower extremity kinematics and electromyographic (EMG) data. The test setup is shown schematically in Figure 1.

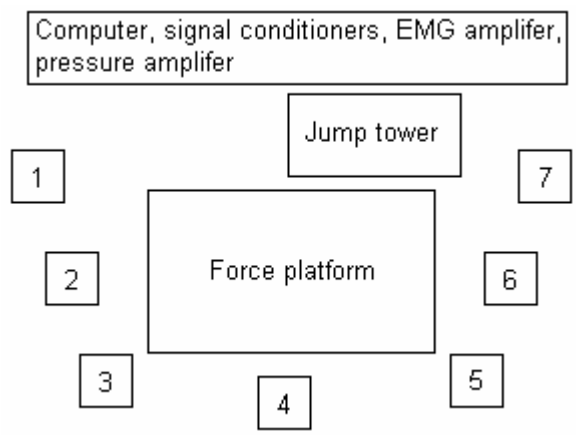

Figure 1: Schematic of testing area including positions of the seven cameras.

Kinematics data were collected using seven digital-cameras, and a high-speed motion analysis system. Surface electromyography was recorded using a 12-channel amplifier and electrode system to determine the level of muscle contraction/activation. Ground reaction forces were collected through the force platforms, which were centered in a large testing area and placed flush with the ground.

Fifteen retro reflective markers, three on each of the trunk, pelvis, thigh, leg and foot segments, were placed on the right limb to determine the three-dimensional lower extremity angles at the trunk, hip, knee and ankle instead of the left limbs. The markers on each segment formed a triad and were fixed to a rigid structure that was strapped to the segment. Figure 2 presents a schematic of the marker set.

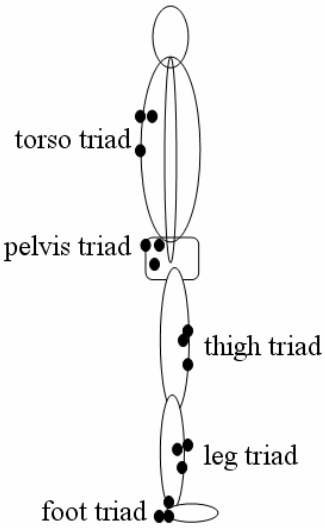

Figure 2: Marker position for kinematics data.

The major emphasis of this test was to re-evaluate an existing ankle brace design (Aircast) and to evaluate the modified ankle brace design (Form fit) and commercially available laced footwear. Seven of the eight previously participated adult males from ages 28 to 68 years took part again in the third test. They went through the same procedures as described in the prior tests. In this test, each subject completed four footwear conditions (first condition is with no protective foot wear) at the targeted landing velocity. The test conditions are tabulated in Table 1. In all cases only Bates boots [1] were used, limiting the complexity of the test. Also, there was no rubber layer covering the force plate in this test.

Table 1: Four test conditions in Test 3.

Test Condition	Foot Wear
C1	No Ankle Brace
C2	Internal Laced Footwear
C3	Form Fit Ankle Brace
C4	Air Cast Ankle Brace

The first condition (C1) in this test was the same as the condition with high landing speed and Bates boots in Test 2. It was used to confirm consistent results. Condition 2 (C2) involved wearing a laced "ankle brace" prior to wearing the Bates boots, as shown in Figure 3. Condition 3 (C3) involved wearing a modified form fit ankle brace, as shown in Figures 4 and 5. This form fit ankle brace is based on a commercial ankle brace. It was modified by

elongating the bottom strap using Velcro connections, thus allowing the ankle brace to be worn on the exterior of the Bates boots. The form fit padding consists of many hexagon-shaped rubber tubes; lay out in a honeycomb matrix that is more compliant to the human ankle contour. The two padding are secured to the foot by the Velcro straps at the top. Condition 4 (C4) involved wearing the Aircast ankle brace that was used in Test 1. The ankle brace is shown in Figure 6.

Figure 3: Internal laced footwear.

Figure 4: Modified "form fit" ankle brace.

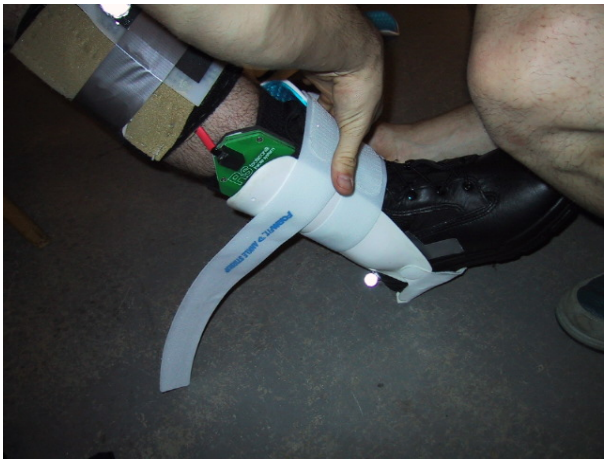

Figure 5: Participant putting on form fit ankle brace.

The four test conditions were carried out in a balanced order design to alleviate a possible order effect. In addition, subjects were allowed sufficient time between conditions or trials to prevent any fatigue effects. Before each jump, each participant stood on the force plate for initial weight calibration. The general experiment setup including the newly placed retro reflective markers, jump platform, and high speed cameras are shown in Figure 7.

Figure 6: Participant putting on Aircast ankle brace.

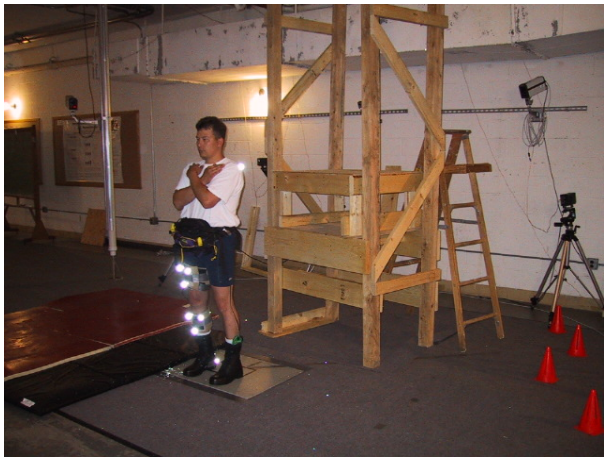

Figure 7: A subject standing on the platform for calibration.

Test Results

The actual landing velocities derived from the kinematic data are presented in Table 2. The nominal landing velocity was designated as 5.18 m/s and a platform was constructed that was sufficiently high to generate such a velocity, all the subjects reached about 90% of the expected velocity.

Table 3 shows the first and second peaks of the vertical ground reaction force, measured in body weight ratio (BWR), with the times where the peaks occurred. The times where peak 1 and peak 2 occurred were very similar

for all of the footwear conditions. The peak 2 value of C4 was greater than other footwears but not significantly. Figure 8 presents the vertical ground reaction force for an exemplar subject in each of the footwear condition.

Table 2: Mean landing velocities (and standard deviations) in all conditions.

Condition	Speed (m/s)	Percent of Required Speed
1	4.75 (0.24)	91.7
2	4.64 (0.29)	89.6
3	4.62 (0.29)	89.2
4	4.65 (0.18)	89.8

Table 3: Mean values (and standard deviations) of times to peaks and peaks of vertical force (BWR) for different conditions.

Parameter	C1	C2	C3	C4
Time to Peak 1	0.014 (0.005)	0.013 (0.005)	0.013 (0.005)	0.012 (0.005)
Peak 1	7.50 (1.67)	7.20 (1.36)	7.25 (1.08)	6.46 (1.33)
Time to Peak 2	0.033 (0.008)	0.032 (0.008)	0.035 (0.009)	0.032 (0.008)
Peak 2	8.88 (2.65)	8.41 (1.63)	8.67 (2.06)	9.08 (1.82)

C1 – No foot wear; C2 – Laced ankle brace; C3 – Form fit ankle brace; C4 – Aircast ankle brace.

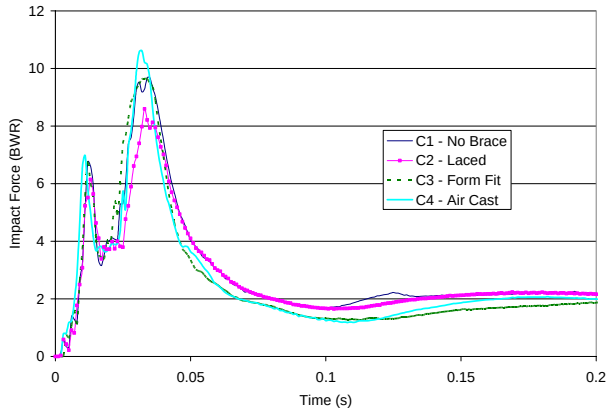

Figure 8: Vertical ground reaction force for an exemplar subject in each of the footwear condition.

The ankle rotations (dorsiflexion and xversion) are tabulated in Table 4, and the dorsiflexion and xversion profiles of an exemplar subject in various footwear conditions are shown in Figures 9 and 10. It can be seen from the kinematics of the PLF that there was a definite reduction in dorsiflexion when foot wears were applied. Using a stiffer foot wear (Aircast), the reduction in both dorsiflexion and inversion becomes prominent. It is important to note that flexibility in the dorsiflexion rotation can attenuate the amount of impact directed to the foot sole. Thus the use of a protective foot wear for paratroopers should reduce the inversion and has minimal affect on the dorsiflexion. In addition, although subjective, but through surveying the participants it was found that the laced and Form fit ankle braces provided better comfort when worn than the Aircast ankle braces; and it was also observed that it would take significantly longer time to put on the internal laced footwear than Form fit ankle braces.

Table 4: Peak dorsiflexion and xversion angle (degree) of right ankle of an exemplar subject in various foot wear conditions.

Condition	Dorsiflexion	Inversion
C1	75.0	27.0
C2	45.0	25.0
C3	45.0	25.0
C4	27.0	15.0

C1 – No foot wear; C2 – Laced ankle brace; C3 – Form fit ankle brace; C4 – Aircast ankle brace.

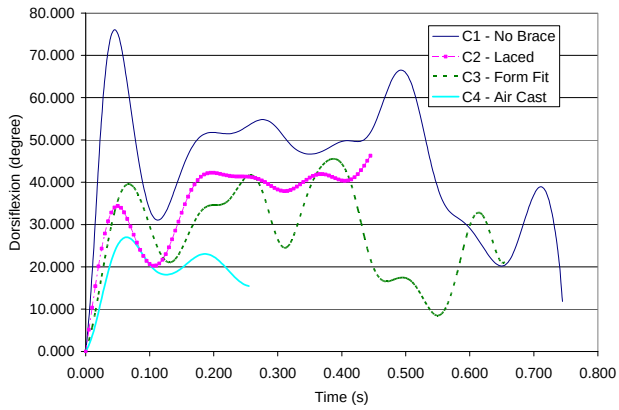

Figure 9: Dorsiflexion profiles of right ankle of an exemplar subject in various foot wear conditions.

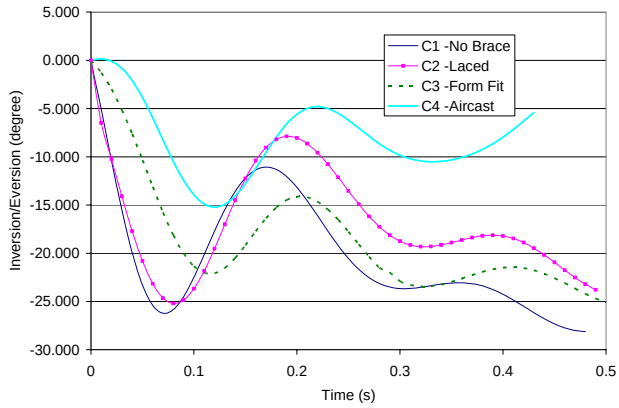

Figure 10: Xversion profiles of right ankle of an exemplar subject in various foot wear conditions.

FINITE ELEMENT MODELING

Model Development

A biomechanical model of the paratrooper was generated from the GEBOD (Generator of Body Data) [2] program. GEBOD is a program from Armstrong Laboratory of Wright-Patterson AFB. GEBOD is a public domain database and can be used to generate human models of various sizes. GEBOD generates the anatomical parts of the human body as rigid ellipsoids with proper dimensions and inertial properties. These ellipsoids are connected together by joints to represent the whole human body, as shown in Figure 11. The model used in this study corresponded to a 50th percentile male and was properly configured to represent the body position of a paratrooper prior to landing. The 50th percentile male has a height of 1.75 m and a weight of 76 kg. The joint properties from GEBOD were modified to more accurately represent the nonlinear, rate dependent characteristics of the human joints based on recent data reported in the literature.

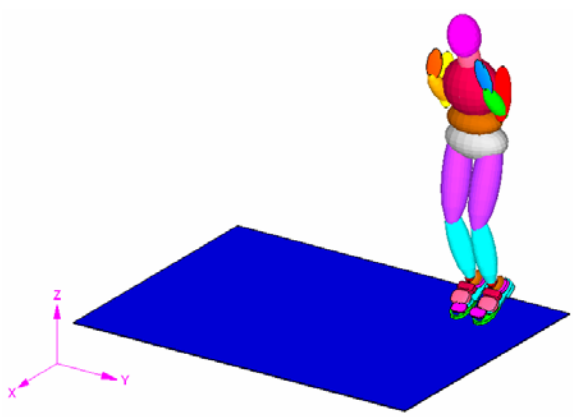

Figure 11: The paratrooper model generated from GEBOD.

LS-INGRID [3] was used to convert the GEBOD model database from an LS-INGRID input file to an LS-DYNA [4] input file. LS-INGRID is a pre-processor for LS-DYNA. The explicit commercial FE code, LS-DYNA has been widely used to solve three-dimensional nonlinear dynamic problems especially in the automotive industry. Its advanced capabilities for modeling impact include sophisticated contact features, large deformation, and an extensive library of nonlinear material models. In addition, the boots, ground, and the ankle braces were added to the GEBOD model using HyperMesh [5]. HyperMesh is a pre- and post-processor for FE solvers and is very effective for model generation and manipulation, especially for contact definition.

One of the major emphases of the current research is to model the effect of external ankle brace usage. Thus in the paratrooper model, the two ankle braces designs were incorporated based on their general geometry and stiffnesses. They are both modeled using elastic material properties. Figures 12 to 15 show the Formfit (C3) and Aircast (C4) ankle braces, respectively. The boundary and loading conditions were the same as the paratrooper model in the previous paper [1].

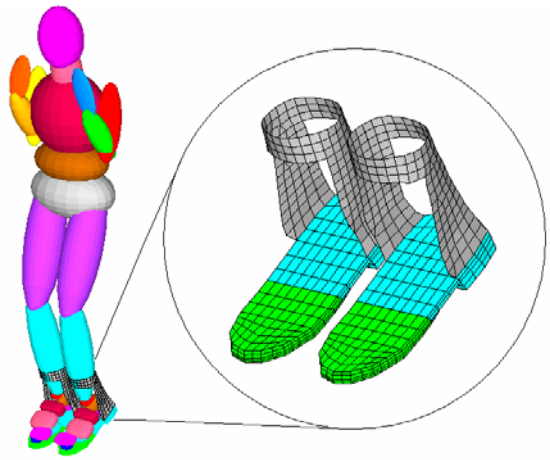

Figure 12: Model of Formfit (C3) ankle brace.

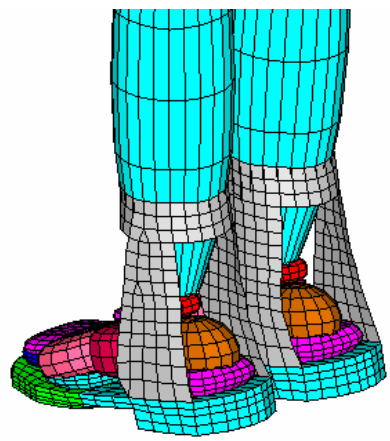

Figure 13: Rear side of the Formfit ankle brace model.

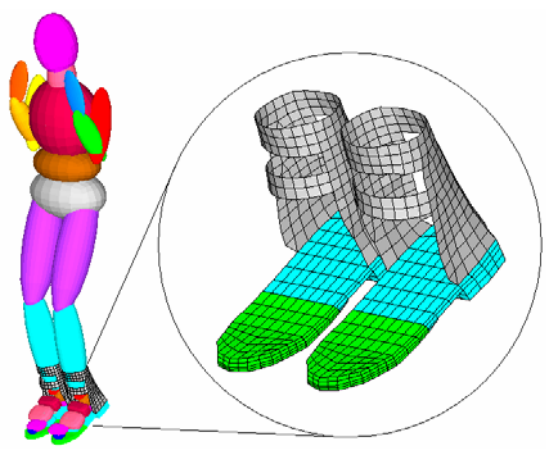

Figure 14: Model of Aircast (C4) ankle brace.

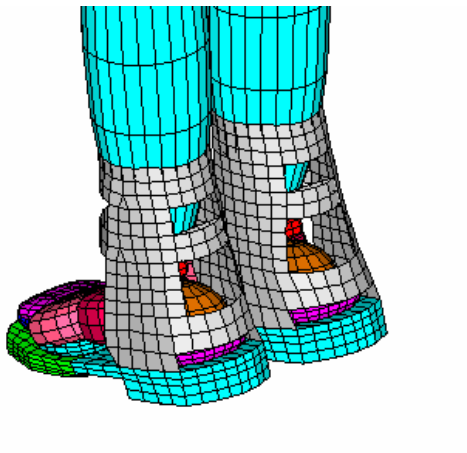

Figure 15: Rear side of the Aircast ankle brace model.

SIMULATION RESULTS

Form Fit Ankle Brace Simulation (C3)

The paratrooper model incorporated with Form fit ankle brace was simulated under the condition of Test 3 with a landing velocity of 4.6 m/s. Figure 16 shows the general kinematics of the paratroopers after the knee touch-down, and the animation is similar to those observed in Test 3.

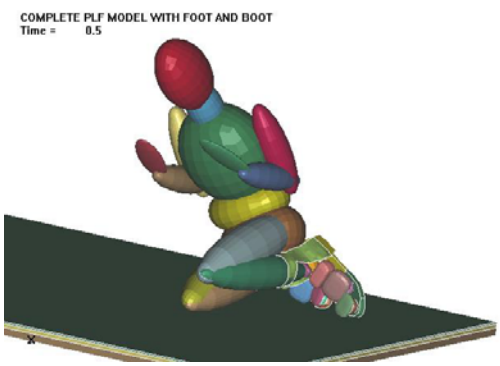

Figure 16: Simulation of PLF with Form fit ankle braces.

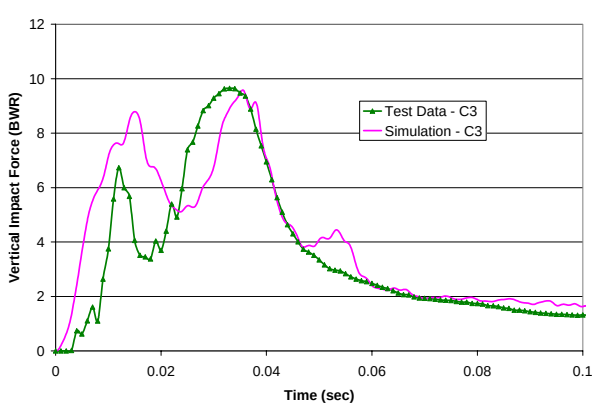

Figure 17: Comparison of vertical impact force (BWR) between test data with simulation with Form fit ankle braces.

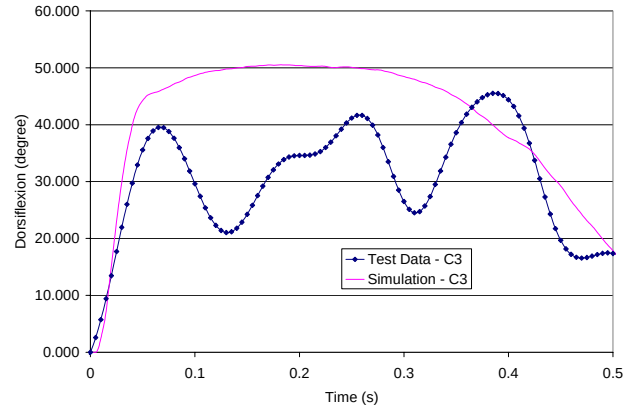

Figure 18: Comparison of dorsiflexion between test data with simulation with Form fit ankle braces.

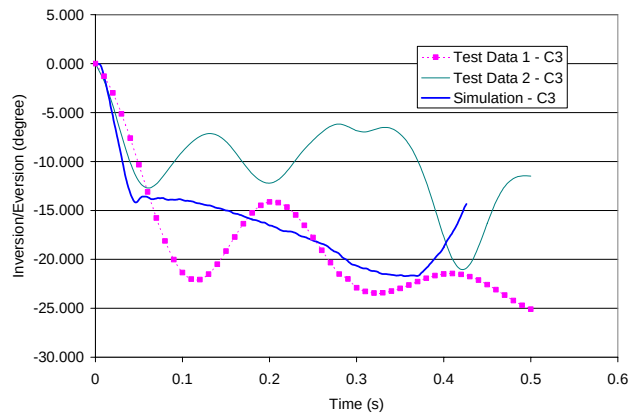

Figure 19: Comparison of inversion rotation between test data with simulation with Form fit ankle braces.

The results are in agreement with the test data in general. The simulation shows that the Form fit ankle brace has no significant influence on the vertical impact force, as shown in Figure 17; and the peak value is closely matched. Both the dorsiflexion and inversion rotations are shown to have similar trends in Figures 18 and 19, despite the oscillations in the test data.

Aircast Ankle Brace Simulation (C4)

The paratrooper model incorporated with Aircast ankle brace was simulated under the condition of Test 3 with a landing velocity of 4.6 m/s. Figure 20 shows the general kinematics of the paratroopers after the knee touch-down, and the animation is similar to those observed in Test 3.

The simulation results show discrepancies from the test data. The simulation predicts that the first peak of the vertical impact force would increase due to the stiff Aircast ankle brace constraint in dorsiflexion, as shown in Figure 21; however the test data showed an increase in the second peak instead. Figures 22 illustrate that the model

predicts a sharper decrease in dorsiflexion after the toe touch-down, which agrees with the trend in the test data despite the difference in magnitude and lack of test data after 0.28 second. Lastly the model predicts a lower inversion rotation as shown in Figure 23 which is still slightly larger than the available test data.

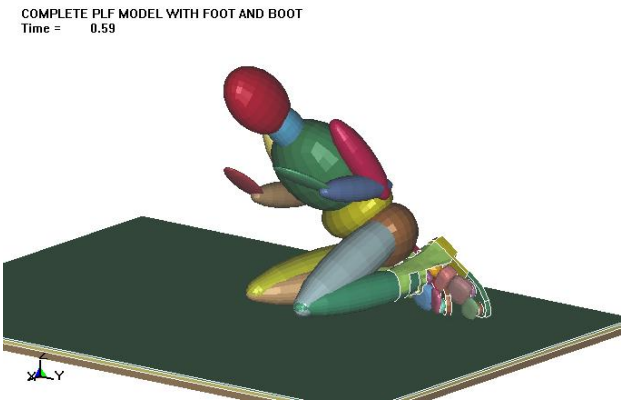

Figure 20: Simulation of PLF with Aircast ankle braces.

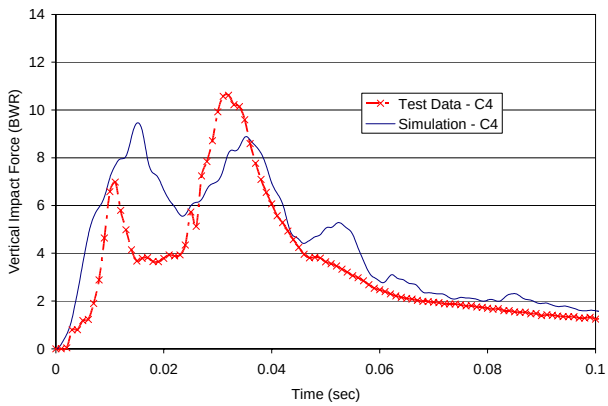

Figure 21: Comparison of vertical impact force (BWR) between test data with simulation with Aircast ankle braces.

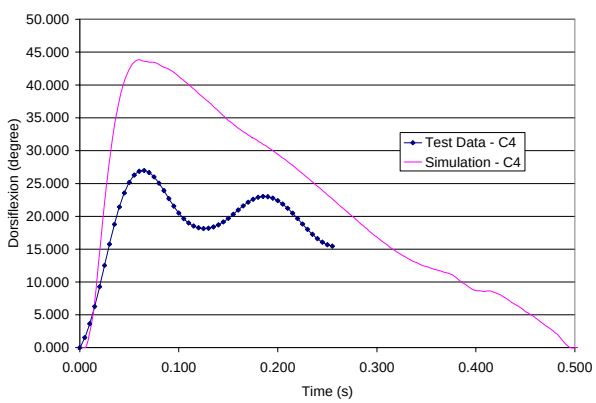

Figure 22: Comparison of dorsiflexion between test data with simulation with Aircast ankle braces.

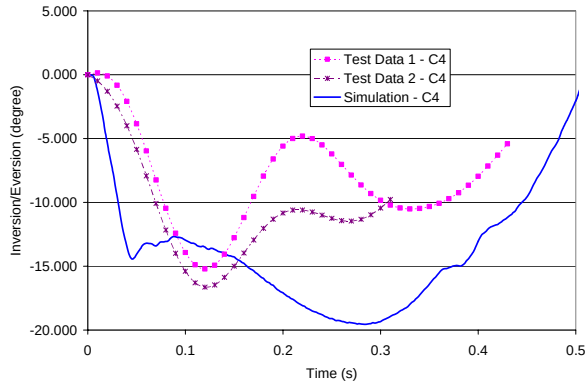

Figure 23: Comparison of inversion rotation between test data with simulation with Aircast ankle braces.

In summary the paratrooper model with Form fit ankle brace was validated through Test 3 data. The model can be used for further design evaluation. While the Aircast ankle brace model agrees to the test data only to a certain degree, it is understood that the source of discrepancy can be contributed to the lack of solid muscles in the ankle/foot complex of the paratrooper model. The Aircast ankle brace has more constraints over the ankle/foot complex. Most of the constraints are related to tightening the ankle brace to the entire foot rather than a distinct Velcro location as in the Form fit ankle brace. Hence the lack of solid muscle support in the paratrooper model causes the Aircast brace to be less stiff than expected. Even with the top portion of the boots modeled, which allows the Aircast brace to be tied throughout the ankle/foot complex, yet the feet are modeled as rigid segments and not a continuous solid representation of flesh. Modeling of the solid muscles such that the mesh is capable of deforming based on the prescribed joint axes and the corresponding material properties can be an ongoing research and development of the currently validated paratrooper model.

PARAMETRIC STUIDES USING THE PARATROOPER MODEL

To better understand the mechanism of PLFs and to identify possible improvements in the area of injury prevention and reduction, parametric studies were performed using the paratrooper model to study the effect of various parameters on impact force and ankle rotations.

Three parameters are selected for the parametric studies. They include the landing velocity, load carriage carried by the paratrooper, and terrain slope. Table 5 shows the parameters and their corresponding ranges to be studied, and Table 6 shows the injury tolerances for the lower extremity and the references.

Table 5: Various parameters and their corresponding ranges used in parametric studies.

Parameter	Range
Landing Velocity (m/s)	4.27, 5.18, 6.00
Load Carriage (lb)	0, 45, 90
Terrain Slope (degree)	-10, 0, 10

Table 6: Injury tolerances for the lower extremity and their reference sources.

Body Parts	Criteria	Tolerances	Reference Sources
Ankle	Dorsiflexion (degree)	±45	[6]
Ankle	Inversion (degree)	±35	[7]
Ankle/Foot	Axial Force (N)	6,750	[8]
Femur	Axial Force (N)	10,000	[9]
Pelvis	Side Impact (N)	9,500	[10]

Effects on Landing Speed

Landing velocity was investigated first, because it is the single most influential parameter. It was believed that landing speed lower than 4 m/s would be unlikely to result in injury, thus 6 m/s was selected on top of the two speeds (4.27 and 5.18 m/s) studied previously. The model setup is based on the C1 condition. The variations of the vertical impact force, dorsiflexion and inversion due to differences in landing speed are shown in Figures 24 to 26, and tabulated in Table 7.

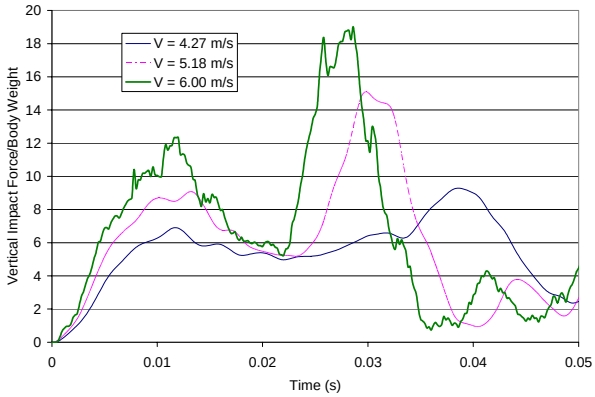

Figure 24: Vertical impact force (BWR) of paratrooper resulted from various landing speeds.

It was observed that as the landing speed increases, the rotation rate for dorsiflexion increases causing the vertical force peak to be reached faster. Also, both the dorsiflexion and inversion angles increases as the landing speed increases, due to a greater inertia from the upper body. The results show that a landing speed beyond 6 m/s would cause injury in the ankle/foot due to high vertical impact force.

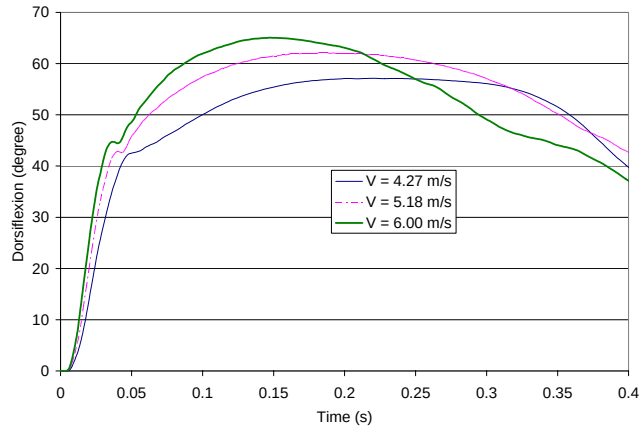

Figure 25: Dorsiflexion of paratrooper resulted from various landing speeds.

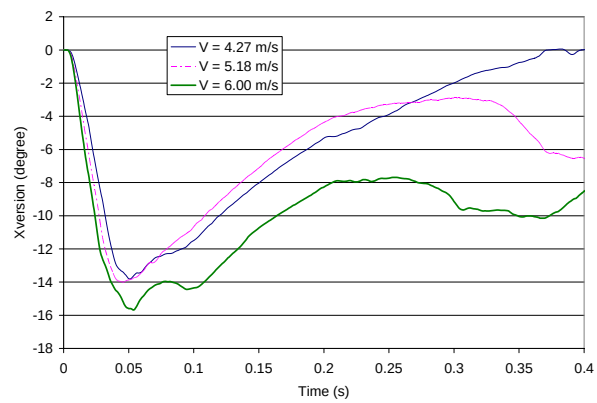

Figure 26: Inversion of paratrooper resulted from various landing speeds.

Table 7: Peak values of various parameters resulted from different landing speeds.

Vo	**4.27 m/s**	**5.18 m/s**	**6.00 m/s**
Impact Force (BWR)	9.17	15.05	19.00
Dorsiflexion (degree)	57.1	62.1	65.1
Inversion (degree)	-13.8	-14.0	-15.7

Effects on Load Carriage

Load carriage carried by the paratrooper was analyzed ranging from 0 to 90 lb. The model setup was based on the C1 conditions. The landing speed was set to be 5.18 m/s to observe results from higher landing speed. For the load carriage carried by the paratrooper, the extra load

was distributed over the thorax portion of the paratrooper model. In addition, based on the kinematics data, the muscle activation levels of the ankle complex, knee and hip joints were adjusted to 30% and 40% for 45 lb and 90 lb load carriage, respectively. This is to accommodate the additional muscle contraction while significant weight is applied on the paratrooper. The resulting vertical impact force, dorsiflexion and inversion of right foot due to the added mass are shown in Figures 27 to 29, and tabulated in Table 8.

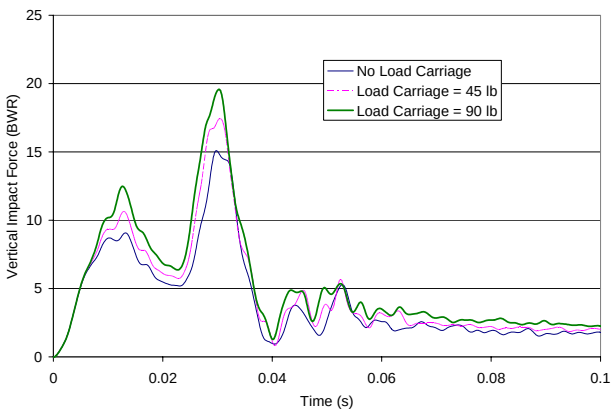

Figure 27: Vertical impact force (BWR) of paratrooper resulted from various carriage loads.

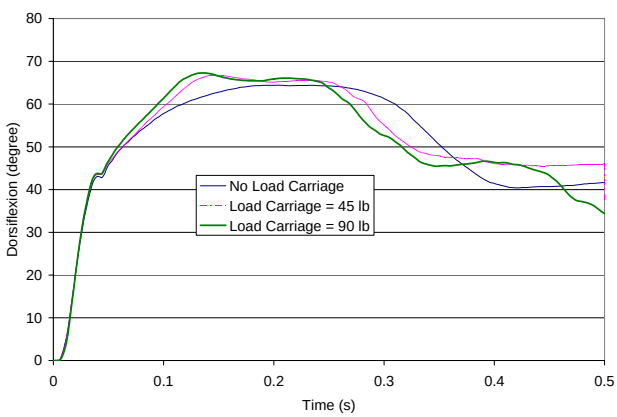

Figure 28: Right-foot dorsiflexion of paratrooper resulted from various carriage loads.

It was observed that as the carriage load increases, the vertical force peak increases. The dorsiflexion angle also increases as the load increases, due to a greater inertia from the upper body. However, as the load further increases; the muscle activation is also increasing, which will affect the kinematics. Thus the inversion angle increases when the load increases from 0 lb to 45 lb, but decreases when the load is increased to 90 lb. From the simulation it was also observed that the kinematics was altered slightly from the intended PLF due to the new mass distribution, which is important since it suggests that any paratrooper carrying load carriage heavier than 90 lb would have difficulty to perform PLF without significantly activate the muscle contraction. These parametric results provide an insight on practical load carriage limitation.

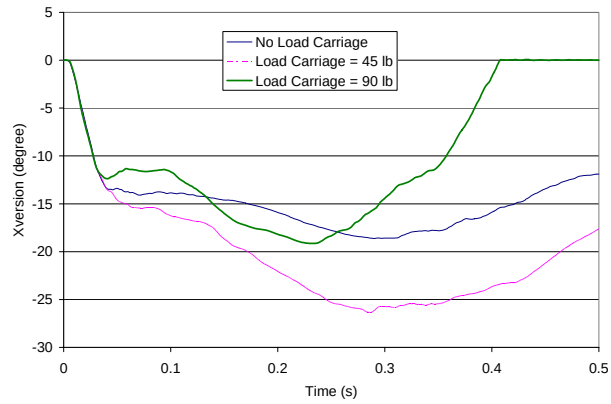

Figure 29: Right-foot inversion of paratrooper resulted from various carriage loads.

Table 8: Peak values of various parameters resulted from different carriage load.

Load Carriage	0 lb	45 lb	90 lb
Impact Force (BWR)	15.09	17.44	19.57
Dorsiflexion (degree)	64.43	66.71	67.28
Inversion (degree)	-18.62	-26.33	-19.13

Effects on Terrain Slope

For the terrain slope variation, the ground was tilted 10 degrees forward and backward. Again the model setup was based on the C1 condition. The landing speed was set to be 5.18 m/s. The variations of the vertical impact force, dorsiflexion and inversion of right foot due to the added mass are shown in Figures 30 to 32.

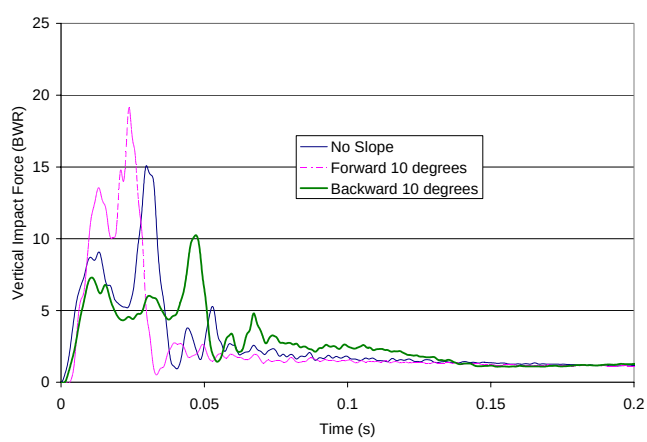

Figure 30: Vertical impact force (BWR) of paratrooper resulted from various terrain slopes.

It was observed that forward incline would increase the peak vertical impact force, and backward incline would decrease it. This is because with the forward incline, the boot heel would impact the ground before the ankles apply the resistance through rotational moment, and this is opposite for the backward incline case. Both the dorsiflexion and xversion angles are very sensitive to forth and aft terrain slope, because the terrain slope can directly alter the contact engagement between the boots and the ground, causing a change in the rotation.

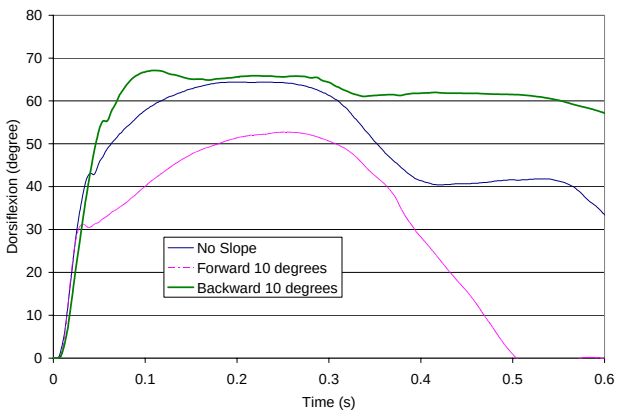

Figure 31: Right-foot dorsiflexion of paratrooper resulted from various terrain slopes.

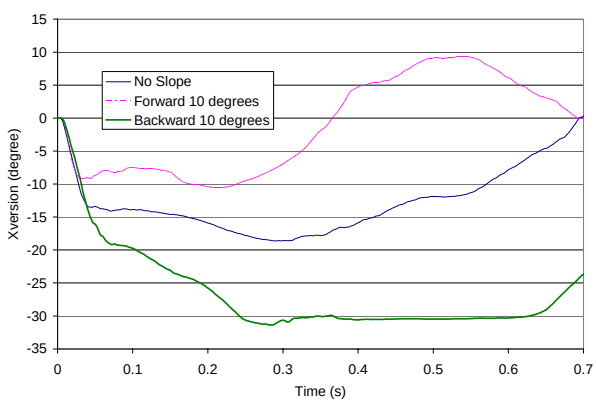

Figure 32: Right-foot inversion of paratrooper resulted from various terrain slopes.

To further understand the complete scope of terrain slope variation, both left- and right- inclined slope were modeled to analyze the injury mechanism in addition to the forth- and backward-inclined terrain. With the same landing conditions, the ground was tilted 10 degrees to the left or to the right in the two cases, and the orientation of the left-inclined is shown in Figures 33.

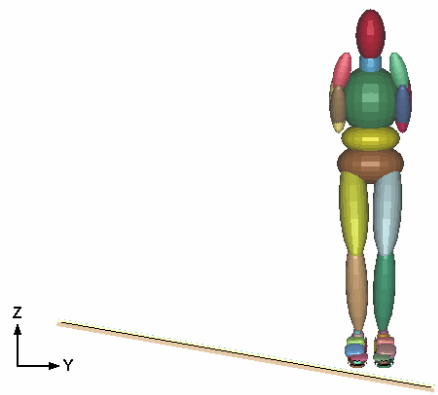

Figure 33: PLF model with left-inclined terrain slope of 10 degrees.

The variations of the vertical impact force, dorsiflexion and inversion of right foot due to the various terrain slopes are shown in Figures 34 to 36, and all of the peak values from the four cases are tabulated in Table 8.

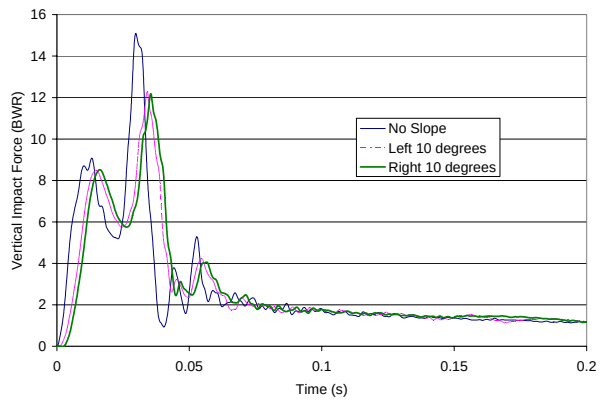

Figure 34: Vertical impact force (BWR) from various terrain slopes at V_o = 5.18 m/s.

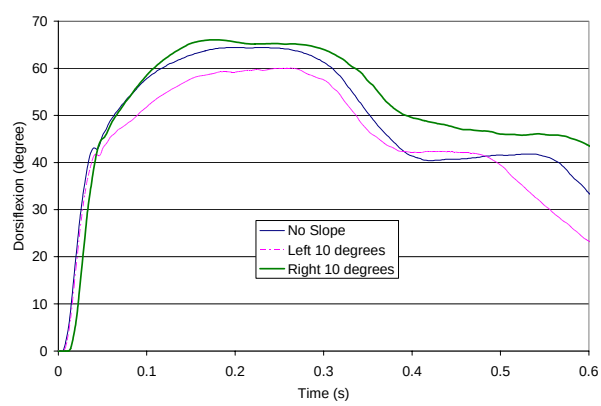

Figure 35: Right-foot dorsiflexion from various terrain slopes at V_o = 5.18 m/s.

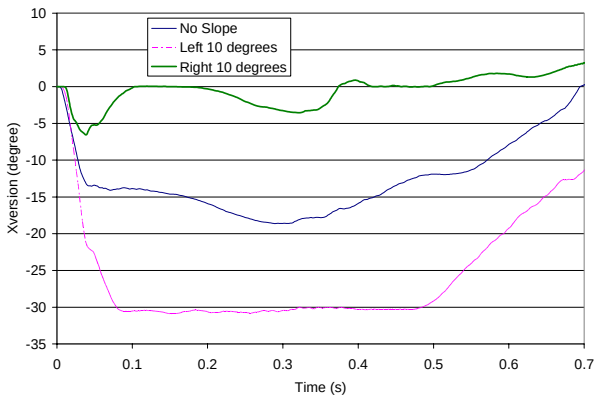

Figure 36: Right-foot inversion from various terrain slopes at V_o = 5.18 m/s.

The results show that both the left- and right-inclined terrains would decrease the peak vertical impact force. This is because with the side incline, only one boot heel would impact the ground initially. The dorsiflexion angle is not sensitive to the side incline; however the xversion angle is very sensitive to the terrain slope. This is because the terrain slope can directly alter the engagement between the boots and the ground in the xversion axis, thus significantly increase or decrease the inversion. Figure 6-15 shows that when the PLF is against the slope (left-inclined), the inversion reaches 30 degrees, the preliminary angle limit set in the PLF model. Figures 37 and 38 show the kinematics of the fall on the left- or right-inclined terrain.

Table 8: Peak values of various parameters resulted from different terrain slopes.

Slope Orientation	No Slope	For-ward	Back-ward	Left-inclined	Right-inclined
Impact Force (BWR)	15.09	19.13	10.24	12.27	12.18
Dorsiflexion (degree)	64.43	52.75	67.11	60.06	66.05
Inversion (degree)	-18.62	10.52	-31.36	-30.88	-6.55

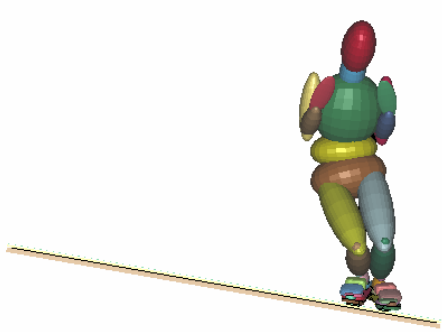

Figure 37: Kinematics of the PLF on a left-inclined terrain.

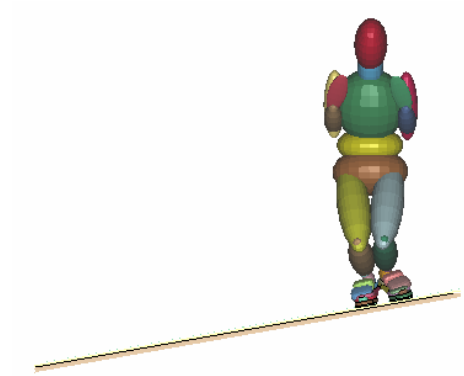

Figure 38: Kinematics of the PLF on a right-inclined terrain.

CONCLUSIONS

A combined approach of laboratory testing and finite element modeling was used to investigate the mechanical behavior of the paratrooper during the PLF. Kinematics and impact forces of the paratrooper of two landing velocities were quantified in the tests. The predicted landing events and impact forces were in close agreement with the test results, especially during the early landing phase. The later stage of the PLF involves mostly active motions such as roll over that are difficult to predict. However, most injuries occur during the early phase of the PLF, the paratrooper model provides a valuable tool for injury and parametric analyses.

The PLF is a pre-programmed event by a displacement perturbation method. The pre-programmed non-reflex muscle action and multi-joint motion during the early phase of impact is important in peak attenuation. Smaller impact force for the PLFs in comparison to stand-up landings is a result of the intention to roll and thus increases the range of body and joint motion in landing during the ground-roll. This ground roll motion serves to minimize the vertical impact force imposed on the ankle/foot complex, only part of the initial kinetic energy of landing is absorbed during the initial stage of landing, while the remaining is transformed into angular kinetic energy and absorbed later.

The ankle/foot complex was developed based on data available in the literature. The Army boot soles were also incorporated in the model and validated through drop test and flexion test. The lower extremity was validated by cadaver test, and the full paratrooper model was validated from the three tests performed at UMass – Amherst. Each test was designed to target specific validation studies. The model was validated based on the vertical impact force, dorsiflexion and inversion rotations. The simulation results show good agreement with test data in the kinematics of the overall PLF process.

The validated model was then used to perform parametric studies with varying landing speeds, carriage loads, terrain slopes and roughness. The effects of landing speed and carriage load were obtained through parametric studies, and the results were compared with the injury criteria to identify critical landing speed and carriage load for paratrooper injuries due to high impact forces.

The terrain slope and roughness were also examined in the parametric studies. The results show that the source of injury is the inversion rotation rather than impact force. The model provides the overall kinematics of the paratrooper under such conditions which are difficult to test physically due to the likelihood of inflecting injury to the paratroopers.

The model was also used to evaluate a couple of protective devices. The results show that as the stiffness of the ankle brace increases, the dorsiflexion and inversion rotation will decrease. The key observation is the balance of dorsiflexion and inversion since the lower dorsiflexion results in higher impact force.

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