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LOWER LIMB RESPONSE TO IMPACT IN SIMULATED MICROGRAVITY AND 1G

DISSERTATION

Presented in Partial Fulfillment of the Requirements for the Degree Doctor of Philosophy in the Graduate School of The Ohio State University

By

Susan Elizabeth D’Andrea, M.S.

* * * * *

The Ohio State University, 1998

Dissertation Committee:
Dr. Brian L. Davis, Advisor
Dr. Alan Litsky
Dr. William Notz

Approved by

Advisor
Graduate Program in Biomedical Engineering
ABSTRACT

The body undergoes a wide variety of physiological adaptations when exposed to an environment of microgravity. One of the major alterations in the musculoskeletal system during weightlessness is the loss of bone mass. The ability of the body to readapt to the Earth's gravitational field after prolonged exposure to microgravity is remarkable, especially when considering the extended periods of time currently spent in space. However, there is still much to learn regarding the adaptive processes in space and their responsible mechanisms. Although many different countermeasures have been used to combat space flight-induced osteoporosis, none have been successful. It has been contended that in order to maintain the integrity of the skeleton, a component of loading must be present. In space, however, skeletal loading on the body is virtually eliminated and reflected in the fact that the body, and especially the lower extremities, lose bone mass. Current exercises protocols used in space to combat bone loss are not effective in providing the necessary axial loading. Therefore, activities which contain a component of high impact force must be investigated as a possible solution.

This study proposes that jumping exercises, which are known to impart high impact loads to the body, will help to combat space flight-induced osteoporosis. The underlying hypothesis is that as the impact load and loading rate increases, so too does the internal strain in the bone. Previous investigations have determined the osteogenic threshold of strain but none have been successful in relating this level to the external forces during activity. Therefore, this study attempted to determine the relationship between the external forces and the internal bone strain during jumping activities.

A zero gravity simulator was constructed and jumping exercises were performed at four different gravity levels in the simulator and in 1G. Twelve subjects, who represented the anthropometric characteristics of the current U.S. astronauts, were recruited for the study. Four of the subjects were instrumented with a calcaneal strain...
transducer in order to assess the level of strain during jumping. Ground reaction forces, tibial and calcaneal accelerations, and ankle and knee joint angles were recorded during jumping. Three different landings were performed at four zero gravity simulator tension levels and in 1G in order to obtain a wide response in the measured parameters. Results obtained in these experiments were compared to drop test data on cadaveric feet. A rheological model of the hind foot, including the fat pad and the calcaneal bone, was also developed in order to predict the strains in the bone and the effectiveness of the heel pad in absorbing the shock transient following high impact forces.

Results showed that strains in the calcaneus were significantly higher in jumping than the proposed osteogenic threshold. However, no clear relationship between the strain and the external ground reaction forces were seen. Differences existed in the level of the force, loading rate and accelerations between the zero gravity simulator conditions and 1G suggesting that the mechanics of jumping is altered in microgravity. Differences were also found for different landing type with the flat-footed landing producing the highest magnitudes in the variables. Comparison of experimental variables to the cadaveric data showed higher magnitudes for the in vivo strains with no apparent differences in acceleration data. The rheological model was successful in predicting bone strains akin to those found in vivo for similar peak forces. The model also showed a highly correlated relationship between force and strain.

The results of this investigation are important in light of the fact that they were able to show that the strain magnitude in the calcaneus exceeded the osteogenic level regardless of either tension level or jump type. This suggests that jumping may be a viable exercise for preventing bone loss in space. It may also prove to be a more cost and time effective alternative to exercise countermeasures in space.
Dedicated to my husband,
James M. Losito
ACKNOWLEDGMENTS

"Some people come into our lives and quickly go. Some stay for a while and leave footprints on our heart and we are never, ever the same."

Anonymous

I came to Ohio three and one half years ago planning on obtaining my doctoral degree but not really knowing what to expect. After completing my classes and candidacy exams, I moved to Cleveland to begin my research. For me, the Cleveland Clinic was a great academic experience. Not only was I able to increase my knowledge in the field of biomechanics, I was exposed to many different aspects of the biomedical engineering discipline. I found myself involved in projects which I would have never thought I'd be working on and learning new concepts which will be useful to me in future work. The staff and personnel at the Cleveland Clinic have been friendly and helpful, always taking the time to assist me. I am grateful for the time I have spent here and the experiences I have had.

My greatest thanks goes to my advisor, Brian Davis, who took me on as one of his students under the recommendation of a mutual friend, without ever meeting me or knowing what I was capable of. He continually pushed me to achieve higher goals and never stopped believing in me, even when I couldn't believe in myself. I would also like to thank my committee members at Ohio State, Dr. Alan Litsky and Dr. William Notz, for their continual support and encouragement throughout this project.

This project could not have been completed without the contributions of my co-workers. It would be difficult to find a better collection of people. Not only did I gain from their individual expertise, but they made working in the lab fun. I would like to thank Dr. Amy Courtney, Ari Levine, David Lord, Tom Lundin, Tiffany Orlando, Tak
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On a more personal note, I thank my parents for always encouraging me to achieve higher goals and to never give up. I don’t think I would have been able to get where I am today without the values which they have instilled in me. Last, but certainly not least, this dissertation belongs to my husband as much as to me. He has constantly supported me and make sacrifices so I could achieve this goal. He has always been there for me, encouraging me to continue even when I was ready to quit, making me smile and being my best friend. To Jim, I give my deepest love and gratitude.
VITA

PERSONAL
1965 Born, New Haven, Connecticut
1995 Married, James M. Losito

EDUCATION
1988 Bachelor of Science, Mechanical Engineering
University of Connecticut, Storrs, Connecticut
1990 Master of Science, Bioengineering
Clemson University, Clemson, South Carolina
Advisor: Christopher L. Vaughan, Ph.D.
1998 Doctor of Philosophy, Biomedical Engineering
The Ohio State University, Columbus, Ohio
Advisor: Brian L. Davis

HONORS
1997 American Society of Biomechanics Graduate Student Grant-In-Aid.
Mathematical Modeling of Load Transmission Through the Calcaneus

PROFESSIONAL EXPERIENCE
1988- Research Assistant
1990 Department of Bioengineering
Clemson University, Clemson, South Carolina
1991- Senior Biomechanical Engineer
1993 The Motion Laboratory
Miami Children's Hospital, Miami, Florida
1993- Research Assistant
1994 Biomechanics Laboratory, Department of Industrial Engineering
The University of Miami, Coral Gables, Florida
1994- Graduate Research Associate
1998 Biomedical Engineering Center, Ohio State University, Columbus, Ohio
Biomedical Engineering, The Cleveland Clinic Foundation, Cleveland, Ohio

FIELD OF STUDY Biomechanical Engineering
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CHAPTER 1

INTRODUCTION

Over three and a half decades ago Yuri Gagarin became the first human in space. Since this inaugural flight, more than two hundred and eighty nine individuals have experienced conditions of microgravity on almost two hundred space flights (Nicogossian, et al., 1994). In fact, space exploration has become an increasingly important objective for the thirteen countries of the European Space Agency, the National Space Development Agency of Japan, the Canadian Space Agency and the National Aeronautics and Space Administration, NASA (Tipton and Hargens, 1996). With continued use of the Soviet Mir Spacestation and the development of the International Space Station, space explorers now have the opportunity to remain in a weightless environment for periods longer than a year. These events, however, have not been without serious physiological consequences.

Although a vast amount of knowledge has been accumulated regarding changes the body undergoes in space, further research is needed to fully understand the adaptive processes and their responsible mechanisms at both the system and cellular levels. Equally uncertain are the measures necessary to eliminate the deleterious effects of space flight.

The human body has evolved in accordance with the presence of gravity which influences growth, development, structure, function, orientation and movement. Without this pervasive force, the body experiences many physiologic changes involving the cardiovascular, neurological, endocrine and musculoskeletal systems. In the musculoskeletal system, the absence of gravity eliminates the necessity of muscles to work against a constant force. This coupled with the fact that the use of the leg musculature for movement and locomotion is almost nonexistent results in decrements in muscle size and
strength, increased muscle fatigue, diminished neuromuscular efficiency and abnormal reflex behavior. This phenomenon had been noted more so in the anti-gravity muscles, in particular the lower extremity extensors.

One of the most profound effects of long duration space flight is the change in bone metabolism. Progressive and continual skeletal changes in space can result in hypercalcemia, kidney stones, lengthy recovery of lost bone after flight with the possibility of irreversible bone loss, increased fracture potential and calcification of soft tissue (Schneider et al., 1994). Calcium balance becomes increasingly negative throughout the duration of the flight and bone mineral density declines. Bone loss has been noted predominantly in the lower extremities while the upper extremity bones show a tendency toward increased mass. Bone losses of approximately 1.15% per month have been reported in the lumbar spine, proximal femur, and the femoral neck (Organov et al., 1997; Shackelford et al., 1997). The calcaneus alone has been reported to lose as much as 17% of its preflight mass after only twenty four days (Kakurin, 1972). Complicating the severity of bone loss in space is the ability of astronauts to recover bone mass after returning to Earth's gravity. Trends suggest that mineral recovery is gradual and takes approximately the same amount of time as the loss (Vogel and Whittle, 1976); however, it is uncertain whether full recovery is possible.

A variety of techniques have been proposed and/or used to alleviate the negative effects of microgravity. These include: exercise, lower body negative pressure, fluid loading, pharmacological therapy and the possible use of artificial gravity. Exercise has been used extensively to combat physiological adaptations in space. Evidence of the importance of exercise in space was demonstrated on the ninety-six day Soviet-26-Salut-6 mission in which no exercise was performed during the initial twenty five days of flight (Georgiyeski et al., 1980). The physical deconditioning after this time period was manifested by increased heart rate, elevated arterial pressure, reduced stroke volume and an inability to complete a five minute exercise regime on the cycle ergometer. Subsequent days included exercise and endurance test and, by the end of the flight, vital statistics returned to pre-flight levels. In fact, after a four week mission without exercise, astronauts will lose perceptible amounts of upper body strength and locomotor function.
After twelve weeks, most crew members will be unable to walk upon their return to earth and will have detectable decreases in bone mineral density (American College of Sports Medicine, 1996). These changes cause no difficulty in space but such losses must be controlled if crews are to function on earth after extended periods in weightlessness.

While exercise modalities used in space have been relatively effective in preventing cardiovascular deconditioning, they have not been successful in preventing bone mineral loss. This may be due in part to a lack of high impact forces on the stationary cycle, rowing ergometer or the treadmill which are most often used during space flight. Researchers have suggested that high loading rates and impact forces are critical for reducing space flight induced osteoporosis (Cavanagh et al., 1992). Numerous animal investigations support this hypothesis. Rubin and Lanyon (1987) explored the concept of strain as the osteoregulating influence on bone. In vivo experiments on the sheep and turkey ulnae showed that strain distribution, strain rate and strain magnitude were indeed osteogenic stimuli. Skeletal integrity was preserved by applying a strain equal to 1000 microstrain (0.1% strain) for at least four cycles per day. This research suggests that when strain magnitude is too low, there will be no adaptive response regardless of the number of loading cycles; however, with sufficient strain, relatively few cycles are required to stimulate bone deposition.

1.1 Statement of the Problem

Because bone is a dynamic tissue which responds to its mechanical environment, in the absence of gravity bone resorption will occur. Although overall bone loss may be relatively low, weight bearing bones are more severely affected. Based on calcium balance studies, it has been predicted that a 5% loss in total body calcium after one year in space reflects a 25% decreased in mineralization of the lower extremities (Whedon, 1984). This is equivalent to a forty year loss in post-menopausal women (Courtney et al., 1995). Additionally, in bones such as the calcaneus which are composed predominantly of trabecular bone, the effect of weightlessness may be magnified because of the increased metabolic activity in the structure. A significant decrease in an important weight bearing bone such as the calcaneus can have serious implications on its strength and function.
Since bone loss in space can have serious consequences for astronauts returning to earth, more effective countermeasures must be developed to combat space flight induced osteoporosis. This cannot be accomplished, however, without a complete understanding of the mechanisms of osteogenic stimuli and how they can be implemented during space travel. In order to do so, a relationship between the external forces acting on the body and the internal strains must be developed. To date, no such relationship exists. Additionally, the type of activity which will engender the level of strain necessary to maintain skeletal integrity must be elucidated to prescribe exercise protocols in space.

1.2 Purpose

The importance of direct physical longitudinal stress has been emphasized for the maintenance of skeletal integrity. Taking this into consideration, activities which impart high impact forces to the body may provide a viable means to supply the body with substantial bone strains and strain rates which in turn will prevent bone loss in the calcaneus. It is therefore proposed that a jumping exercise protocol in microgravity will help minimize bone demineralization. Large dynamic loads and high loading rate have been found for jumping activities on earth. This coupled with the fact that the calcaneus is directly impacted during loading, suggests jumping may be an optimal activity to explore the relationship between high impact force and bone strain.

To address this hypothesis, the lower limb response to jumping exercises in earth's gravity and simulated microgravity will be investigated. Three distinct methods will be used to determine effect of high impact loads on the body. They are: (a) the design and implementation of a zero gravity simulator, (b) the comparison of jumping exercises in simulated microgravity and 1G, and (c) the development of a mathematical model to predict the response of the lower extremity to impulsive loads. The experimental trials will provide an understanding of the ground reaction forces, load transmission and internal calcaneal bone strain experienced during jumping. These measured values will contribute to the input values of the mathematical model and be used to validate the efficacy of the model. The overall goals of this research are (a) to ascertain the relationship between the external ground reaction forces and the loading rates and internal strain in the calcaneus, (b) to determine the factors which significantly affect the ground reaction forces during
jumping, (c) to understand the mechanism of load transmission through the hindfoot, and (d) to show that a mathematical model is a valid means of estimating strain and load transmission through the calcaneus. This research will provide insight into countermeasures which will effectively decrease the bone demineralization which occurs in space.

1.3 Specific Aims and Hypotheses

**Specific Aim 1:** To determine external impact loads and loading rates, calcaneal strain, tibial and calcaneal acceleration, jump landing, jump height, and ankle and knee angles during jumping exercises in simulated microgravity.

**Hypothesis 1:** The magnitude of the external ground reaction force is directly proportional to the tension level in the gravity replacement system and the jump height.

**Hypothesis 2:** The magnitude of the impact force is directly related to the angle of the knee and ankle angle at landing during jumping exercises.

**Hypothesis 3:** Calcaneal and tibial acceleration increase with increasing impact force. These values indicate the level of shock absorption during impact.

**Hypothesis 4:** Peak loads in jumping are dependent upon the type of landing.

**Hypothesis 5:** Loading rate is highly correlated to the peak load in jumping.

**Specific Aim 2:** To compare jumping exercise data (impact loads, loading rate, shock attenuation, calcaneal strain) in simulated microgravity and 1G.

**Hypothesis 1:** The levels of force, loading rate, and calcaneal strain obtained during jumping exercises in 1G can be achieved in simulated microgravity.
Hypothesis 2: No significant difference between tibial and calcaneal accelerations or shock attenuation are evident between zero gravity jumping and 1G jumping.

Specific Aim 3: To establish a relationship between external ground reaction forces and loading rates and the internal strain rate and strain magnitude.

Hypothesis 1: Bone strain increase non-linearly with increasing external force due to the viscoelastic nature of the tissue surrounding the calcaneus and their role in shock absorption during jumping.

Hypothesis 2: Peak external forces can be used to predict internal bone strains.

Hypothesis 3: Since the mechanical properties of the plantar skin and heel pad vary with the rate at which they are loaded, the loading rate increases as the soft tissue becomes stiffer creating an environment with decreased shock absorption. With decreased shock absorption, bone strains become greater. Therefore, as the loading rate increase, the calcaneal strain increases non-linearly.

Hypothesis 4: External loading rates can be used to predict internal bone strains.

Hypothesis 5: Because strain rate highly correlated with the strain, the strain rate will have a similar relationship to the applied force and loading rate as the strain.

Hypothesis 6: The magnitude of strain in the calcaneus will reach values greater than or equal to the 0.1% strain - the threshold of strain prescribed by Rubin and Lanyon (1987) for the maintenance of skeletal integrity.

Specific Aim 4: To compare impact data from cadaveric drop test and in vivo jumping exercises in 1G.
Hypothesis 1: The calcaneal strain *in vivo* is significantly different from strain measured in situ due to the action of the muscles, ligaments and soft tissue.

Hypothesis 2: No significant difference is evident between *in vivo* and in situ accelerations.

Hypothesis 3: The relationships between peak external impact forces and internal bone strains in the cadaveric specimens will predict lower calcaneal strain than *in vivo* data.

Hypothesis 4: Cadaveric measurements of tibial strains can be used to predict tibial strains in 1G and simulated microgravity.

*Specific Aim 5:* To develop a model which can accurately predict the response of calcaneal bone to high impact forces.

Hypothesis 1: A mathematical model has the ability to simulate a physiological response to high dynamic loads.

Hypothesis 2: Shock absorption and calcaneal bone strain predicted by the model is similar to values measure in simulated microgravity and 1G.

Hypothesis 3: The relationship between peak external force and bone strain in the model is similar to the association derived from experimental data for jumping in 1G and simulated microgravity.
CHAPTER 2

BACKGROUND

2.1 Bone

Bone is a complex, dynamic tissue which functions as a specialized framework for
the body, provides protection for internal organs, functions as a mineral reservoir for the
body, serves as the site for hemopoiesis and plays an important role during movement due
to the action of the muscles on the skeleton. Inorganic mineral makes up 70% of bone,
organic matrix and cells 25%, with the remainder predominately water. The organic
matrix of bone consists primarily of type I collagen, similar to that found in skin and
tendon. The basic unit of type I collagen, the tropocollagen molecule, is a triple helix of
polypeptide chains. The molecules are aligned in parallel to produce a collagen fibril. In
bone, these fibrils are grouped together in fibers according to the structure of the
particular tissue - woven or lamellar bone, for example. The mineral portion of bone is
hydroxyapatite [Ca_{10}(PO_4)_6(OH)_2]. Crystals of hydroxyapatite are small, elongated and
hexagonal in shape and conform to the orientation of the crystal fibers. Bone
hydroxyapatite differs from the naturally occurring apatite crystals in that it contains
several impurities including fluoride, magnesium and sodium. This crystalline phase of
bone is an important mineral reservoir for the body containing 99% of the body's total
calcium, 85% of its phosphorus and 66% of its magnesium. While bone cells are critical
to the structure and function of bone, they comprise only 2% of the organic tissue’s
constituents.

Bone is capable of regulating its mass and architecture in order to meet two critical
and competing responsibilities: structural and metabolic. The balance between these
functions is controlled by the deposition and resorption of the tissue through
osteoregulatory cells - osteoclasts, osteoblasts and osteocytes. These cells mediate the remodeling for structural function and regulate the body's metabolic needs for calcium and other minerals. In order to fully understand this adaptive mechanism, this review will include an examination of the architecture, physiology, mechanical properties as well as the remodeling process and the stimuli necessary for bone growth.

2.1.1 The Physiology and Architecture of Bone

2.1.1.1 Gross Anatomy

Bones are generally classified in five categories according to their shape: long, short, flat, irregular and sesamoid. Long tubular bones contain a central diaphysis of cortical bone which surrounds the medullary cavity (Figure 2.1). This central core is lined by the endosteum, a thin, largely cellular layer of connective tissue. The epiphyses are found at the end of long bones. They articulate with other bones and are lined by hyaline articular cartilage. Between the diaphysis and epiphyses are the metaphyses, an area of extensive bone remodeling during development. Surrounding the bone is the tough periosteum which is well supplied with blood vessels and nerves, some of which enter the bone.

The adult has two distinct types of arrangement of bony substance which can be classified according to their porosity. Cortical bone is a dense, solid material which is strong and resistant to bending. The thickness of cortical bone varies according to the specific bone and its mechanical function. Cancellous (also called trabecular or spongy) bone is present at the epiphyseal and metaphyseal areas in long bones and within the cortical bone of flat and short bones. Cancellous bone is very porous and composed of a three-dimensional lattice of plates and columns called trabeculae. The trabeculae orient themselves in the principal direction of the forces which are applied to the bone. Bone with a volume fraction of over 70% is classified as cortical and less than 70%, cancellous (Gibson, 1985). However, porosity is not the only distinguishing factor. Trabecular bone can be differentiated from cortical bone by its architecture.
2.1.1.2 Histology

At the microscopic level, bone is arranged in distinct patterns. The predominant unit of structure in adult cortical bone is the haversian system or osteon (Figure 2.2). Each osteon contains a central, vascular haversian canal surrounded by concentrically arranged lamellae of bone. The haversian system runs parallel to the long axis of the bone contributing to its ability to effectively resist axial forces. Located between adjacent lamellae are the lacunae - small cavities which contain osteocytes. All lacunae are interconnected in the haversian system by tiny canals called canaliculi. The protoplasmic processes of osteocytes in adjacent lacunae enter the canaliculi and make contact for the passage of nutrients and waste from one cell to the next. Blood and lymph vessels and
nerves enter the haversian system from large vessels located on the surface of the bone (in the vascular layer of the periosseum) or through the marrow. Blood vessels from within each of these sources enter the canaliculi through Volkmann’s canals which run at right angles to the osteon. Circumferential lamellae just beneath the periosteaum line the shaft of long bones.

The organization of cancellous bone can be described as an interconnected network of rod and plate-like trabeculae (Carter and Hayes, 1976). Lamellae run parallel to the trabeculae and are oriented in the direction of principle forces. Individual trabeculae are interspersed by marrow which supplies the bone with nutrients and means for waste exchange. Similar to the cortical bone, trabecular bone contains a network of lacunae and canaliculi for the intercommunication of the osteocytes.

Figure 2.2: The Haversian System (Spence and Mason, 1987, pg. 132)
A sophisticated network of osteoregulatory cells are responsible for the constant remodeling of the skeleton. The osteoblasts synthesize and mineralize the extracellular matrix. They are connected via gap junctions to form a continuous blanket of cells on the surface of the bone. As the osteoblast continues to mineralize, it may become engulfed in its own matrix and remain within lacunae becoming a part of the osteocyte network. Osteocytes which project through the canaliculi provide an ideal conduction pathway for chemical and electrical communication within the bone matrix (Curtis et al., 1985). Osteoclasts are bone macrophages which migrate over the bone surface creating irregular cavities or Howship's lacunae. Their function is specialized for bone resorption. The interaction between these three bone cells for the remodeling of bone is highly regulated by the presence of numerous hormones.

2.1.2 Mechanical Properties of Bone

As long, hollow tubes which derive maximal strength with minimal weight, bones are well suited for their structural role (Hayes and Gerhart, 1985). The ultimate strength of bone approaches that of cast iron, it has the capacity to absorb and release twice as much energy as wood and yet its weight is only two thirds that of steel. Equally important are the facts that both cortical and cancellous morphology is strategically arranged to evenly distribute functional stresses (Lanyon, 1973) and that bone adapts to its functional environment to maintain a structure of approximately constant stiffness. Both cortical and trabecular bone contribute to the overall properties of whole bone; however, individually, their characteristics vary greatly due to differences in architecture and porosity.

Cortical bone has been classified as an anisotropic or transversely isotropic material. For normal compressive or tensile loading, the elastic modulus is approximately 17.0 GPa in the longitudinal direction, 11.5 GPa in the transverse direction and 3.3 GPa in shear (Reilly and Burstein, 1975). The mechanical strength of cortical bone is derived from its composite nature of haversian, circumferential and interstitial lamellae. Strength of cortical bone is also highly dependent on the direction of loading as well as the type of loading (Table 2.1). These data clearly indicate that cortical bone is strongest in compression. This suggests that it is well suited for the compressive loading of daily
activities. Since the yield strength of cortical bone is approximately equal to the ultimate strength, bone which is loaded close to its yield point may also be close to fracture. Therefore, bone can undergo great deformation (or strain) before fracture occurs.

<table>
<thead>
<tr>
<th>Loading Direction</th>
<th>Loading Type</th>
<th>Ultimate Strength</th>
</tr>
</thead>
<tbody>
<tr>
<td>Longitudinal</td>
<td>Tension</td>
<td>133 Mpa</td>
</tr>
<tr>
<td></td>
<td>Compression</td>
<td>193 MPa</td>
</tr>
<tr>
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<td>Tension</td>
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<tr>
<td></td>
<td>Compression</td>
<td>133 MPa</td>
</tr>
<tr>
<td>Shear</td>
<td>Shear</td>
<td>68 MPa</td>
</tr>
</tbody>
</table>

Table 2.1: Ultimate Strength of Human Femoral Cortical Bone
(adapted from Reilly and Burstein, 1975)

Strain, a dimensionless unit, is expressed in bone physiology literature as microstrain ($10^6$). The yield strain of cortical bone is approximately 7,000 microstrain (0.7% change in length) with the ultimate strain reaching up to 15,000 microstrain (Carter et al., 1981). These high magnitudes are possible due to the arrangement of parallel osteons in the bone which will "slip" relative to one another rather than break. The lamellae of each osteon behave in a similar manner allowing for a more elastic system. Mechanical properties of cortical bone are sensitive to the rate at which the strain is applied. Higher strain rates increase the elastic modulus, yield and ultimate strengths, and for very high strain rates cause the bone to become more brittle. In day to day activities (0.1 strain per second or less), however, there is no noticeable appreciation in modulus and the ductility increases with increasing strain (Keaveny and Hayes, 1993).

Trabecular bone is best described as an open-celled porous foam (Gibson, 1985) and its mechanical properties vary with density, architecture, anatomical site and age. Evans and King (1961) were the first to recognize that the physical properties of trabecular bone were a function of its location in the human body. Even within the proximal tibia, the material properties of the trabecular bone can vary by up to two orders
of magnitude (Goldstein et al., 1993). Both the density and the architecture of the bone are major determinants in overall strength; however, both of these factors depend on the anatomic site and, to some extent, age.

Both modulus and strength of cortical bone are related to the apparent density. The precise relationship between these variables, however, remains an open question. Galante et al., (1970) reported a linear relationship between strength and density while Carter and Hayes (1977) reported a squared power dependence. Carter and Hayes also showed a cubic relationship between modulus and apparent density. Current sentiment favors a squared relationship for both modulus and strength in trabecular bone alone (Goldstein, 1987). These differing reports, however, confirm that factors other than density contribute significantly to trabecular bone properties.

For trabecular bone, there is a relationship between compressive strength and the elastic modulus (Goldstein et al., 1983). Therefore, stiffer trabecular bone is also proportionally stronger. This suggests that strain is the controlling factor for failure. This is supported by evidence that trabecular bone yields at strain in the range of 1 to 4% with a very weak dependence on density (Brown and Ferguson, 1980; Turner, 1989). While the tensile behavior of trabecular bone differs from its compressive behavior, studies are inconclusive regarding the relative magnitude of compressive or tensile strengths. However, the clinical implications of these relationships between strength, modulus and apparent density are clear: bone can easily adjust its strength and stiffness by changes in bone density which are accomplished through remodeling and subtle changes in density will result in large changes in strength and modulus.

2.2 Mechanical Stimuli for Bone Growth

"Form follows function." This concept put forth by Wolff in 1892 is probably one of the most quoted in the orthopaedics community. Wolff saw that the balance of bone remodeling can be affected by mechanical function and that bone tissue has the capacity to adapt to its functional environment such that its morphology is optimized for its mechanical demand. The exact nature of this relationship and the specific mechanisms or mechanical stimuli which control the cellular processes of bone remodeling, however, are
not yet fully understood (Rubin et al., 1990). Therefore, the ultimate goal of researchers in this field must be the clarification of the transduction pathways that relate mechanical stimuli to the integrated remodeling response of bone. To do so, investigations have focused on answering the following questions:

1. What component of the functional environment is osteoregulating?
2. What magnitude and duration of stimulus are necessary for the maintenance of skeletal integrity?
3. What is the structural objective of bone morphology?

The underlying basis for the concepts developed to answer these questions has been the assumption that strains which are measured during functional activity should indicate what the architecture of the skeleton is trying to accomplish. This is illustrated ideally when considering loading in bone. Axial loads delivered to the bone will be sustained with very small strain. However, it is not the axial component but rather the bending moments which are responsible for over 80% of the strain at the bone surface (Rubin and Lanyon, 1982). Although bending moments will always exist in the structure of bones, they can be significantly reduced by alteration of the longitudinal curvature. However, the orientation of bone does not appear to be directed toward this neutral bending axis and in some cases is directed to increase bending (Rubin, 1984) suggesting that bone curvature acts to accentuate strain rather than limit it.

2.2.1 The Daily Stress Stimulus Theory

Theoretical bone modeling generally considers that bone skeletal integrity is maintained over a certain threshold of stress/strain stimulus. Any stimuli above or below this threshold initiates bone deposition or resorption, respectively. Carter et al., (1987) proposed a relationship to estimate the daily tissue level stress stimulus (φ) from the effective stress magnitude (σ) and the number (n) of daily loading cycles (Equation 2.1):

\[
φ = \left[ \sum_{day} n_i (\sigma_i) \right]^{1/2m}
\]

\[
ρα = \left[ \sum_{day} n_i (\sigma_i)^m \right]^{1/2m}
\]

(2.1, 2.2)
where m is the stress exponent parameter which weights the contribution to the bone tissue stimulus from the number of loading cycles and cyclic stress magnitudes. This relationship suggests that the bone in a state of equilibrium must experience and average daily bone stimulus which is a function of the loading history. Additionally, the average daily stress history (\( \sigma \)) is directly proportional to the bone apparent density (\( \rho \)) assuming a constant stimulus for all bone tissue (Equation 2.2). The relationship between external loading and the relative densities of the bone can be derived from Equation 2.2 based on the assumption that all loads can be scaled to the magnitude of the ground reaction force (Equation 2.3).

\[
\frac{\rho_{s}}{\rho_{i}} = \left[ \frac{\sum_{day} n_{i}(P_{s})^{m}}{\sum_{day} n_{i}(P_{i})^{m}} \right]^{1-2m}
\]  

(2.3)

This equation was further developed by Whalen et al. (1988) to estimate the stress exponent m and to apply the model to human data. By considering hypothetical combinations of body weight and activity level, the value of m was established between 2 and 8 with lower values of m correlating to bone densities of more active individuals and higher values indicative of sedentary lifestyles. Whalen and colleagues applied the model to two previous running studies in which calcaneal bone mineral density was measured demonstrating that the stress magnitudes, or in the case of walking or running, the joint forces, have a greater influence on bone mass than the number of loading cycles. This is consistent with the results from animal studies (Rubin and Lanyon, 1984) and suggests that physical activity can predictably affect bone mass. A more critical implication from this investigation is that exercises with high impact loading components may be beneficial in increasing bone mass.

Beaupre et al. (1990) revised the model of daily stress stimulus (Equation 2.1) eliminating the exponent 2 and using a value of m=4. Using this modified equation, the authors showed the existence of a "normal" range of stimulus for which no appreciable gain or loss of bone is present. For values greater than the normal stress stimulus, bone apposition occurs and conversely, for values less than the normal, bone resorption results.
Other investigations have used the model for daily stress stimulus to study the distribution of bone density in the adult proximal femur and have substituted energy equivalent strain into Equation 2.1 (Beaupre et al., 1990b; Mikic and Carter, 1995). Adams et al. (1997) applied the daily stress stimulus theory to in vivo strain measurements of the adult turkey ulna to assess whether the model accurately discriminated maintenance and adaptation in cortical bone. The daily strain history of the turkey ulna was established by cataloging and counting the wing activities during a 24 hour period. Applying the data to Equation 2.1, the authors found that the daily stress stimulus theory did not effectively predict bone adaptation: some maintenance regimes were associated with a higher daily stress stimulus than some regimes which cause bone formation. The authors attribute this finding to several factors. First, the strain measured at the rosette strain gauge site is not necessarily the peak strain in the bone, an inherent problem in any in vivo strain measurement. Also, the daily stress stimulus theory assumes that the summation of peak magnitude is the determining factor in adaptation. Theoretical and experimental data, however, have shown the existence of a "threshold" in mechanical stimulus which dictates when bone is lost, maintained or increased (Beaupre et al., 1990; Frost, 1983; Rubin and Lanyon, 1987). This study does not invalidate the daily stress stimulus theory but merely suggest that several temporal features of mechanical loading such as number of cycles, frequency, duration and rate may play critical roles in bone adaptation along with the magnitude and number of cycles of the stimulus.

2.2.2 Strain as an Osteogenic Factor

It has been proposed by numerous investigators that bone remodeling is continually influenced by the level and distribution of function strains in the bone. Frost (1983), like many other investigators, has proposed that a feedback mechanism exists in bone that senses the effect of mechanical load and converts it into a signal which will control the adaptive remodeling of the bone. The effective stimulus of this feedback loop is the strain, and more particularly, the minimum effective strain (MES). MES is defined as the minimum level of strain that predicts exactly when and where bone adaptations occur in response to mechanical loading. The magnitude of the MES ranges between 80
and 2,000 microstrain and any strains below this value would not evoke the adaptation mechanisms in the bone. Frost (1983) further states that all strain can be classified in three categories: (a) strain smaller than the MES; (b) strains greater than MES; and (c) MES. The underlying principle behind this proposal is that any strains which exceed the MES would cause adaptive remodeling which would decrease the strain experienced by the bone to a level below the MES. A similar theory, the adaptive hypothesis, is proposed by Rubin (1984).

The effect of altered strain environment was clearly demonstrated in a study evaluating the bone mineral density of the patella (Sievanen et al., 1996). Dual energy X-ray absorptiometry (DXA) of the patella and isometric strength data from a twenty six year old female were analyzed over a three year period: a one year unilateral strength training program followed by an accidental anterior cruciate ligament rupture and two years of rehabilitation. Strain index, as defined by the authors as a function of the maximum isometric force, bone mineral density and the patellar length were assessed during each of these three periods. Loading during the initial year increased the strain index of the patella by 47% percent with no significant effect on the bone mineral density. Immediately following injury, the strain index was drastically reduced (most likely due to the immobilization and lack of muscle activity) resulting in a 25% loss of bone mass. During rehabilitation, strain increased drastically (to 135% compared to the baseline), most likely to initiate bone formation. Increases in bone mass were observed until an apparent balance between strength and bone mineral density was evident. These results are consistent with the nonlinear nature of skeletal response to mechanical stimuli. This study also illustrates that unusual strain state evident in the patella in its compromised state as rehabilitation began served as an osteogenic stimuli to return the bone to its original strain state by increasing bone mass. Additionally, this mechanism suggests that bone adapts to maintain an optimal strain environment.

Using technology developed by Evans (1953), in vivo measurements of bone strains were first investigated on animal models in the 1970's (Cochran, 1972; Lanyon, 1973; Lanyon and Smith, 1970). Based on rosette strain gauges, this technique allowed
for the magnitude and direction of the principle strains to be assessed during physiological activities. Such data are of limited significance since the results are restricted to only a small portion of the overall bone area; however, this technology is the only type currently available which can provide any insight into the response of bone to mechanical deformation in vivo. Measurements of surface strain at different sites have been made on numerous species from man to fish showing similar activity despite differences in species and/or limb configuration (Goodship, 1992) suggesting that the limits of strain are universal to all bones and that strain sensitive cellular response must exist. These studies have provided considerable evidence of the response of bone to perturbations of the normal strain environment.

Rubin and Lanyon (1987) have stated that "...bone remodeling, both locally and throughout the skeleton, is continually influenced by the level and distribution of the functional strains within the bone" and that it is the strain that is primarily responsible for the control of bone mass throughout life. To account for this theory, they have developed the adaptive hypothesis (Figure 2.3). Contrary to the normal interpretation, the ultimate goal of adaptation is not minimizing strain but rather to generate an "optimal" level of strain. When a new activity is introduced, such as jogging or aerobics, skeletal strains are increased beyond the optimal environment. This stimulus initiates bone deposition which subsequently reduces the functional strain. Conversely, with disuse or decreased activity, bone resorption will occur thus increasing the strain state of the particular bone.

2.2.3 Studies by Rubin, Lanyon and Colleagues

A large majority of the work on strains and bone remodeling has been conducted by a group of veterinary scientists. In a series of experiments using various animal models, this group of investigators set forth to determine the functional adaptation of bone to increased stress. In a study involving pigs, a doubling of strain in the radius induced by ulnar ostectomy resulted in hypertrophy which not only restored the cross sectional area of the bone but also the customary strain experienced by it (Goodship et al., 1979). These data suggest a feedback mechanism with displacement control and involves several
pertinent factors including: the number of cycles and their frequency, the amount of strain, strain rate and strain duration. Lanyon and colleagues studied strains in sheep models in several different investigations (Lanyon and Bourn, 1979; Lanyon et al., 1982; O'Connor et al., 1982) determining that the direction and magnitude of the strain were not as critical as the strain distribution and strain rate for the functional adaptation of bone.

![Figure 2.3: The Adaptive Hypothesis for Bone Remodeling (adapted from Rubin, 1984)](image)

The results of these and other studies using osteotomy or tenotomy are limited because of the lack of control over the various components of the mechanical environment which may have an effect on bone homeostasis. The most which can be determined in such experiments is the comparison between the changes in measured parameters of the mechanical environment and the subsequent remodeling. This problem is minimized with the use of external loading (Hert et al., 1969; McDonald et al., 1994); however, the
physiological relevance of applying loads for a limited, arbitrary time period is questionable. In order to isolate and control parameters specific to the mechanical environment of bone, Rubin and Lanyon (1984) developed the functionally isolated ulna preparation to which external loads could be applied in vivo.

In the first set of experiments using this preparation in roosters, the adaptive remodeling response was measured with cyclic loading. Isolated rooster ulnae were either not loaded or subjected to loads of 0.5 Hz, 4, 36, 360 or 1800 times daily for a maximum of 42 days. In the disuse models (no loading), bone loss was noted after the initial two week period in the form of endosteal resorption and intracortical porosis. Bones loaded for only 4 cycles per day remained essentially unchanged with respect to bone mineral content. Thirty six cycles per day resulted in a 33% increase in bone mineral content; however, additional cycles beyond thirty six had no effect. The results clearly demonstrate the response of remodeling activity to the number of load cycles, consistent with the findings of Whalen et al. (1988) and Mikic and Carter (1995). More specifically, relatively few cycles (4) are adequate for the maintenance of skeletal integrity.

In a second investigation on turkey ulnae, Rubin and Lanyon (1985) determined the strain in the bone and the "load applied/strain engendered" relationship. Instrumented with a device similar to that of the rooster, bones were deprived of strains except for the daily loading of 100 consecutive 1 Hz load cycles to produce the desired strain. Animals were sacrificed after an eight week period and the midshaft of the ulnae sectioned and prepared for analysis. Strains below 1,000 microstrain were associated with bone loss, however, values of strain at or above this level showed skeletal maintenance and osteogenic activity, respectively. The results confirm a definite "dose/response" curve for the adaptive stimulus with a threshold value of 0.1% strain. Frost (1983) and Beaupre et al. (1990) also indicated the existence of a threshold level for osteogenic strain. The effect of dynamic loading versus static loading was also investigated by Rubin and Lanyon (1984b) using the turkey ulna model. With strain levels equal, statically loaded bones showed no increase in new surface bone formation whereas the dynamically loaded ulnae
experienced up to a 40% increase in bone area. These support the theories that static loads have no effect on bone homeostasis and are, in fact, ignored as an osteogenic stimulus.

While the studies have concentrated on the osteoregulatory nature of mechanical stimuli in cortical bone, the adaptive response in trabecular bone is not as clear. The majority of studies of trabecular architecture and bone density in relation to functional loading have focused on finite element analysis techniques of the proximal femur (Beaupre et al., 1990b; Carter et al., 1989) and the effects of age-related and disease processes (Parfitt et al., 1983). Experimental studies, on the other hand, have focused on the effects of disuse. Biewener et al. (1996) showed that the trabeculae in marsupial calcanei are aligned with the principle strain direction of the cortical surface. This is consistent with the findings of Lanyon (1974) and supports the trajectorial theory of trabecular organization. Although the trabecular alignment remained the same after eight weeks of disuse, bone volume ratio, trabecular thickness and trabecular number decreased by 35%, 25% and 16%, respectively compared to the contralateral control. A similar study in sheep calcanei (Thomas et al., 1996) resulted in a 29% decrease in trabecular bone volume, thinning of the individual trabeculae, but no decrease in trabecular number. Skerry and Lanyon (1995), also working with sheep models, found a 22% decrease in bone mineral content of the calcaneus after twelve weeks of disuse. Interrupting the disuse by short periods of walking exercises was insufficient to prevent bone loss and peak strains engender at the test site were reduced. These data suggest that the strains experienced during normal quadrupedal locomotion have little effect on bone architecture or, conversely, that the strain distributions and magnitudes need longer periods of time to exert their action.

2.2.4 In Vivo Human Bone Strains

Knowledge of the strains experienced during normal daily activities has direct implications in the study of osteoporotic postmenopausal women, fractures and fracture healing and prosthetic replacement. To date, however, very few investigations have monitored bone strains in humans and the understanding of the mechanical control of bone
remodeling has been derived almost completely from animal models. Likewise, all studies have focused on the tibial cortical bone and not addressed issues in the trabecular bone. Three investigations in the literature have analyzed the strain in the human tibia during walking, running (Lanyon et al., 1975), vigorous activities (Burr et al., 1996) and jumping (Milgrom, 1998). The results from these studies not only have helped to clarify the level of strain which can be elicited in humans but also helped to validate the similarity of human bone strains to those found in animal studies.

In the study by Lanyon and colleagues (1975), a 35 year old male was instrumented with a rosette strain gauge on the anteromedial aspect of the tibial midshaft. Strain traces were recorded for walking and running on a treadmill belt and on the floor, with and without shoes for each condition. Principle strain and principle strain angle were calculated for each condition. Simultaneous filming at 64 frames per second allowed for correlation of the subject's position with strain data. A distinct four phase pattern of strain was determined for walking: two during swing and two during stance. During swing, the first phase occurs at initial swing as the limb begins to progress forward, the next in terminal swing just prior to heel strike. During stance a deformation phase occurs between foot-flat and heel off and just before toe-off. Table 2.2 shows a summary of strain magnitudes and strain rates during walking and running.

There are several significant findings of this study. First, it showed that it is technically possible to bond a rosette strain gauge in vivo and record strain on cortical bone. Secondly, it demonstrated that the strain angle changes without noticeable changes in gait. Results also showed that strain were significantly different in the shod condition compared to barefoot. This suggests that footwear significantly effects the functional environment of the bone. Most importantly, this study established that the strain on the bone during gait consists of a series of discrete events in which bone is deformed, unloaded and then deformed in the opposite direction. This seems to indicate that the muscles play a critical role in the strain developed on the tibia.

Although the study by Lanyon et al. (1975) was the first to establish in vivo bone strains, they did not consider vigorous activities which may result in extreme strain and
lead to possible stress fractures. Burr et al. (1996) hypothesized that the incidence of stress fractures may be caused by momentary high strains or strain rates during vigorous activities. To investigate this hypothesis, rosette strain gauges were implanted in two subjects on the medial tibial cortex at the midshaft. Subjects performed activities similar to military field training exercises which often cause stress fractures in recruits.

Compressive, tensile and shear strains were recorded during: (a) walking, jogging and sprinting on level surface; (b) walking carrying a 17 kg pack; (c) walking and running uphill and downhill; and (d) zigzag running uphill and downhill.

<table>
<thead>
<tr>
<th>Max Strain Rate (x 10^3)</th>
<th>Treadmill Walking</th>
<th>Floor Walking</th>
<th>Treadmill Running</th>
</tr>
</thead>
<tbody>
<tr>
<td>Phase 1 Compressive Strain</td>
<td>-154 to -178</td>
<td>-89 to -113</td>
<td>-220 to -300</td>
</tr>
<tr>
<td>Phase 1 Tensile Strain</td>
<td>73 to 98</td>
<td>41 to 64</td>
<td>124 to 177</td>
</tr>
<tr>
<td>Phase 2 Compressive Strain</td>
<td>-104 to -140</td>
<td>-82 to -224</td>
<td>-273</td>
</tr>
<tr>
<td>Phase 2 Tensile Strain</td>
<td>210 to 254</td>
<td>128 to 337</td>
<td>746 to 847</td>
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<tr>
<td>Phase 3 Compressive Strain</td>
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<td>-308 to -425</td>
<td>N/A</td>
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<td>Phase 3 Tensile Strain</td>
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<td>237 to 388</td>
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</tr>
<tr>
<td>Phase 4 Compressive Strain</td>
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<td>-67 to -113</td>
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</tr>
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<td>Phase 4 Tensile Strain</td>
<td>35 to 171</td>
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</tbody>
</table>

Table 2.2: Compressive and Tensile Strains in Walking and Running. Strain expressed as microstrain, strain rate as microstrain/s. Phase 1 - Heel Strike; Phase 2 - Flat-foot to Heel-off; Phase 3 - Heel-off to Toe-off; Phase 4 - Forward Swing. (Adapted from Lanyon et al., 1975).
Results showed principle compressive strains ranged from -414 microstrain in downhill walking to -1226 microstrain during uphill zigzag running. Similarly, principle compressive strains were between 381 microstrain (walking on a level surface with pack) and 743 microstrain (zigzag uphill running). Compressive strain rates were highest during sprinting and downhill zigzag running while tensile strains reached their peak during sprinting but were also high in downhill running. The most vigorous activities including zigzag running and sprinting showed strains and strain rates two to three times greater than walking on a level surface which confirms the hypothesis of Lanyon et al. (1975). Although previous animal studies have reported strain in excess of 2,000 microstrain, these levels were not attained in the human tibia during strenuous activities. The authors suggest that this may be attributed to the age of the subjects as well as the speed at which the animal studies were performed. Additionally, the authors stress that the maximum strain may not have been measured because of the experimental set up and the fact that strain recordings were taken from a single location on the tibia.

During jumping, tibial strains averaged 0.18% in compression, 0.09% in tension and 0.54% in shear (Milgrom et al., 1998). Peak compressive and tensile strains in the tibia generated during landing from a jump were not significantly different from those found during vigorous military training activities (Burr et al., 1996). However, it is important to note that the shear strain is approximately 3.6 times greater than compressive or tensile strains during jumping.

2.3 Skeletal Unloading

2.3.1 Bed Rest Studies

The adverse effect of prolonged bed rest deconditioning in humans was recognized by Hippocrates who reported the loss of strength and performance after prolonged bed rest and inactivity (Chadwick and Mann, 1950). Today, it is recognized that bed rest causes a loss of hydrostatic pressure in the vasculature below the heart, elimination of longitudinal compression of the spine and long bones of the lower extremities, reduced muscular force on bone and reduced total energy utilization (Convertino, 1997).
Additionally, bed rest contributes significantly to the ability to perform physical work. These factors combined, bed rest provides an ideal means to assess the effect of unloading the skeleton and has been used extensively for this purpose.

Early bed rest studies showed a significant increase in urinary and fecal calcium excretion. Deitrick _et al._ (1948), studied four healthy young males subjected in a seven week bed rest studies which included immobilization with casting from the waist to toes. Calcium excretion increased progressively until week four at which time it leveled off to 2-3 times higher than normal. At the end of the seven weeks, a 14.1 g loss of total body calcium was found. Subsequent bed rest studies have confirmed these results (Donaldson _et al._, 1970; Schneider and McDonald, 1984). Donaldson _et al._ (1970) found a decrease in the elevated calcium excretion after three to four months of bed rest; however, levels remained above the normal range. The total loss of calcium for this study was approximately 0.5% per month with daily negative balance of -200 to -300 mg/day. In the same investigation (Donaldson _et al._, 1970), the loss of bone mass in the radius and the calcaneus was evaluated. Although the radius showed no significant loss, the trabecular bone of the calcaneus lost up to 40% of its mass (almost 5% per month) suggesting that the lower extremities are more susceptible to unloading.

Fecal calcium excretion of calcium can account for 50% of the negative calcium balance after eight weeks of bed rest (Schneider and McDonald, 1974) which suggests a reduction in the intestinal absorption of calcium. More recent studies have confirmed that the calcium absorption decreases approximately 22% from baseline after 17 weeks of bed rest (LeBlanc _et al._, 1995). Bed rest induced decreases in intestinal absorption, however, these decreases alone are not enough to account for the large increases in urinary calcium. The same investigation showed alkaline phosphatase and osteocalcin showed no decrease during bed rest indicating that bone formation is not decreased. In a conflicting report, French and Russian investigators analyzed iliac crest bone biopsy data after 120 days of head-down tilt bed rest to determine if alterations in bone mineral density are the result of
increased resorption or decreased formation, (Vico et al., 1987). No significant change was found in the formation of nonmineralized bone matrix while the mineralization of newly formed bone appeared impaired.

The rate and extent of bone loss and recovery from disuse has been extensively covered in the literature. LeBlanc et al. (1990) found significant losses in the total body, lumbar spine, femoral neck, trochanter, tibia and calcaneus of six normal males after 17 weeks of bed rest. Decreased bone mass values, expressed as a percentage of the baseline, were 1.4, 3.9, 3.6, 4.6, 2.2 and 10.4, respectively. During the six months of ambulation following bed rest, only the calcaneus showed a significantly positive slope (p<0.05) with almost 100% recovery. Bone loss in spinal cord injured patients have also been observed; however, BMD loss during bed rest is significantly less than decrements in lower extremity bone mineral density observed after spinal cord injury (Bloomfield, 1997). Changes in bone mass with prolonged bed rest, exposure to microgravity and following spinal cord injury are summarized in Figure 2.4.

Figure 2.4: Alteration in Bone Mass Due to Skeletal Unloading. Black bars indicate alterations in bone mineral density while gray bars show changes in iliac crest trabecular bone volume. (Adapted from Bloomfield, 1997).
Recovery of bone during re-ambulation following bed rest studies is normally slower than the rate at which it is lost. Cancellous bone mass lost in the immobilized hind limbs of dogs after twelve weeks of casting was eventually regained during twelve weeks of normal weight bearing; however, dogs who were immobilized for longer periods of time did not return to their baseline level of bone mass even after twenty eight weeks of ambulation (Jaworksi and Uhthoff, 1986). Residual deficits were more pronounced in older dogs suggesting alteration in the bone deposition mechanisms in older animals. To prevent rapid loss of bone mineral during bed rest and the possibility of fracture upon return to normal activity, the effect of various therapeutic treatments during bed rest has been examined including strength training exercises, static longitudinal compression, intermittent longitudinal compression, impact loading and lower body negative pressure. Although several studies have shown that exercise may prevent bone losses (Whedon et al., 1949; Vogt et al., 1965; Lynch et al., 1967), the majority of these studies have fallen short of demonstrating any clear potential for bone demineralization. Schneider and McDonald (1984) found that none of the countermeasures mentioned above had significant effects on reducing the mean negative calcium balance after bed rest or preserving the calcaneal bone mineral density.

2.3.2 The Skeleton in Space

The predictions of bone loss due to weightlessness were confirmed as early as 1963 during the Vostoks II and II missions when cosmonauts exhibited increased levels of urinary calcium excretion (Goode and Rambaut, 1985). Apollo and Gemini astronauts exhibited similar findings of negative balance also through increased urine and fecal calcium excretion (Zernicke et al., 1990). In fact, hypercalcuria, a consistent finding in astronauts, appears to be the primary factor contributing to negative calcium balance in space (Wronski and Morey, 1983). Whedon et al. (1977) reported a calcium loss of 140 to 184 mg/day and a phosphorus loss of 220 to 400 mg/day in Skylab crew members. Excretion of hydroxyproline was also increased suggesting an increase in bone resorption.

Because the skeleton is a major reservoir of the body's calcium, alterations in urinary and fecal calcium homeostasis reflect changes in skeletal mass. Bone density
measurements have quantified changes in the radius, ulna and calcaneus. Bone mineral density declined up to 8% in the calcaneus after the Skylab missions while no significant changes were found in the radius or ulna (Zernicke et al., 1990). These data indicate that the weight bearing bones may be more sensitive to demineralization during weightlessness. More specifically, trabecular bone may be particularly susceptible to the effects of space flight. The ability to obtain conclusive evidence regarding the pathogenesis of bone loss in space, however, is difficult without histomorphometric data. Additionally, if such losses compromise the integrity and strength of the bone, the success of future long-duration missions is questionable. Irreversible bone loss, accompanied by age-related osteopenia, can result in increased fractures in astronauts, years after returning from space. Studies of the reversibility of space-flight induced osteoporosis have shown a residual deficit in bone mineral density of the calcaneus in an American astronaut even five years after returning from space (Tilton et al., 1980). Bed rest studies, however, have reported conflicting results (Donaldson et al., 1970).

Because physiological responses during weightlessness differ from those under gravity conditions, it is possible that the remodeling process in bone is altered in space. Since bone formation is preceded by bone resorption, a temporary deficit in skeletal mass occurs between the termination of resorption and the initiation of deposition in the remodeling process. If weightlessness alters the bone cell activity, osteoclastic activity may be enhanced while osteoblastic activity is depressed creating irreversible osteoporotic effects in astronauts. Conversely, if weightlessness induces an increased rate of remodeling, bone lost during resorption would eventually be replaced with no long term consequences (Wronski and Morey, 1983). The mechanism of bone loss in space, however is still unclear. As mentioned previously, the loading environment of bone can significantly effect its mass and may be an effective means to combat osteopenia. Burr et al. (1989) attempted to define the specific alterations in bone tissue kinetics responsible for the geometric bone changes in response to altered strain environments. The data demonstrate that bone's response to altered strain occurs primarily by altering the balance of resorption and deposition on the surfaces which are remodeling. Although intracortical
adaptation may occur, there is no evidence this leads to increased bone mass. Further study is necessary to fully understand the tissue kinetics associated with bone remodeling.

To understand the mechanisms of bone loss and the alterations in skeletal structure and composition after space flight, animals have been sent into orbit for scientific study. On Cosmos flights, rats exhibited a decrease in periosteal bone formation (Wronski and Morey, 1983b), a decline in trabecular bone mass in the tibial and humeral metaphyses and a diminished osteoblastic population (Jee et al., 1983). These data further imply that the main factor contributing to bone loss during space flight is diminished bone formation. Significant changes also occur in the geometry and mechanical properties of rat bone as a consequence of weightlessness. In the cortical bone of the humerus, flexural rigidity, elastic modulus, load at the proportional limit, maximum tensile stress and maximum load were all significantly decreased in flight animals. Similarly, tibial flexural rigidity and maximum load were significantly less. Elastic modulus was 13% less in flight animals compared to control although this difference was not statistically significant. Bone geometry measurements indicated decreased humeral length and tibial diaphyseal cross-sectional area (Shaw et al., 1988). Similar results were found in the trabecular vertebral bone of rats after twelve and a half days in space (Zernicke et al., 1990).

2.4 Countermeasures to Physiological Deconditioning

Countermeasures to the physiological deconditioning in space have focused on simulating Earth-normal physical movements, applying gravity-like forces and stresses to the body and creating interactions with the environment similar to that found in 1G. The most direct approach would be to artificially generate gravity in space. To date, however, this has not been possible technically or economically. Instead, fluid loading, electrostimulation, strength training, penguin suits, lower-body negative pressure devices, anti-gravity suits and numerous exercise programs have been utilized in both the Russian and US space programs to combat cardiovascular and musculoskeletal deconditioning. Additionally, nutritional and pharmacological therapies have been proposed as possible countermeasures although their mechanisms for absorption and action are not fully understood in microgravity.
Since the early 80's emphasis has been placed on the development of more efficient countermeasures and, in particular, on protocols which minimize the time needed for physical training. Historically, ground-based simulations have been the basis for the design of countermeasures employed during space flight; however, these methods do not fully replicate the adaptations that occur in microgravity. Life science missions have provided valuable scientific information regarding the validation of the current countermeasures as well as the identification and prediction of the deconditioning in space. These findings have had a profound effect on the development and use of countermeasures which facilitate the readaptation to the Earth's atmosphere although the ideal countermeasures have yet to be determined.

2.4.1 Various Countermeasure Approaches

Cosmonauts use a specialized, constant-loading suit (the "penguin" suit) for up to eight hours a day during orbit. These suits provide resistance to the arms, legs and torso thorough a series of elastic bands which extend from the shoulders to the waist and from the waist to the legs. The purpose of the penguin suits is to compensate for the absence of gravity by opposing the movement of the antigravity muscles which have been shown to be most affected in weightlessness (Gazenko, 1980). The effectiveness of this treatment, however, is difficult to quantify since the elastic cords are not calibrated and cosmonauts are free to make adjustments to the tension level for comfort.

Lower-body negative pressure (LBNP) devices generally consist of a vacuum suit worn up to the waist. Its purpose is to induce fluid shifts towards the lower extremities and has been used as a means of maintaining cardiovascular fitness as well as assessing orthostatic intolerance upon landing. Fluid loading in combination with LBNP has been implemented to help restore fluid lost during flight. The anti-gravity suit, or the G-suit, is opposite to the LBNP devices in that it applies increased pressure on the lower extremities in order to prevent blood pooling upon returning to Earth.

Electrostimulation has been practiced as a means of conditioning muscle and preventing disuse atrophy without the time consuming constraints of physical exercise. Although results from electrostimulation have shown only moderate increases in muscle
function and loss, these gains may be beneficial in space. (Convertino and Sandler, 1995). Electrostimulation as a countermeasure has been incorporated into the Russian space program but has not received wide acceptance in Western aerospace medicine. Additional instruments for strength training of muscles in space include the capstan-type stretcher and the elastic chest expander. These devices are used to help prevent muscle atrophy and fatigue, especially in the upper extremities. Isokinetic strength training devices have also been utilized in space to increase the number muscles affected by exercise and add another dimension to the exercise through movement (Convertino and Sandler, 1990).

Pharmacological therapies are for cardiac deconditioning, muscular dysfunction and bone mineral loss. Drugs such as nitroglycerin, panangin and riboxin, potassium orotate and GABA analogues have been used on Soviet missions to improve the regulation of blood flow (Atkov and Bednenko, 1992). For bone mineral loss in space, nutritional supplements of calcium and phosphorus have shown promising results in bed rest studies and may prove to be effective in space (Schneider et al., 1994). The use of diphosphonates has also been prescribed as a possible countermeasure to bone loss because of its effectiveness in hindering bone turnover (Schneider and MacDonald, 1984). Additional drug therapies that include the use of anabolic drugs have been prescribed for muscle atrophy (Egginton, 1987; Sullivan et al., 1985).

2.4.2 Exercise in Space

The most widely accepted countermeasure to physical deconditioning is exercise. As space flight duration has significantly increased over the past decade, so have the time and intensity of exercise in space. On the first long duration flight, Soyuz-9 cosmonauts were required to exercise for two 1-hour periods per day. Skylab missions also required astronauts to exercise but for shorter periods of time than the Soviets. On the three successive Skylab missions, as exercise periods increased from 0.5 to 1.0 to 1.5 hours per day respectively, post-flight improvements in leg strength, orthostatic intolerance, cardiac output and stroke volume and recovery time were noted even though the later flights were of longer duration (Henry et al., 1977; Thorton and Rummel, 1977). Currently, long duration exercise regimes on the Mir Space Station are divided into three phases: initial
adaptation phase, after the first month and for most of the duration of the flight, and two
months prior to return. Cosmonaut's preferences for particular equipment are taken into
consideration but heart rate should reach an intensity level of 160 to 180 beats per minutes
for one hour, twice daily (Nicogossian et al., 1994). Several strength training session are
also included in the daily routine.

It is difficult to assess the effect of exercise in space. To begin with, exercise
programs are not consistent and outcome is highly dependent on personal effort. It is also
difficult to measure the amount of exercise performed by each crew member especially in
light of the fact that not all activity is reported (Convertino, 1990). Experimental
conditions are less than optimal during missions which often result in experiments which
are not repeatable and data which are difficult to analyze. More importantly, since some
type of exercise is now used on all flights and should not be eliminated due to
consequences to the astronauts, studies of long duration in space without exercise cannot
be conducted. Current exercise in space is focused on aerobic conditioning, primarily as a
means to decrease the cardiovascular deconditioning which occurs in space. Three
aerobic exercise modalities are currently used on Russian and American missions. These
are the treadmill, the rowing machine and the cycle ergometer.

Due to power concerns, weight limitations and vibration issues, a passive treadmill
is used in space. This treadmill belt consists of links of aluminum plates which are moved
as the person walks upon them. A special harness system is used with the treadmill to
tether the astronaut to the apparatus and to pull the subject downward with a force
supposedly close to body weight on Earth. Although the use of the treadmill has been
beneficial, several drawbacks are associated with this device. To begin with, a force
directed down and to the rear is required to start and maintain the action of the treadmill.
This results in abnormal gait where forefoot strike is prevalent with no heel contact as
opposed to the heel-toe pattern in 1G. In addition, the restraint system is not adjustable
once fitted and it is difficult to measure the load pulling the subject toward the belt. It
may also explain why the double flight phase is not seen during running as the harness is
constantly pulling the astronaut downward (Convertino, 1990). These factors decrease
the efficacy of the treadmill to deliver high forces to the body and thus maintain the integrity of the bone and muscle. Designers of the next generation of treadmills have concentrated on developing active belts with sufficient cushioning that will not disturb the trajectory of the spacecraft (Greenisen and Edgerton, 1994).

Cycle ergometers have been useful tools on space missions because they function in several capacities. This device has been used to assess the alterations in hemodynamic and metabolic responses, work capacity and serve as an additional exercise training modality (Convertino and Sandler, 1995). Cycle ergometers were flown on the initial Skylab missions and continue to be used by Russian cosmonauts in flight. These devices can be pedaled with either the legs or the upper arms and the most recent models include programs modules with variable speed and workload patterns.

It has been suggested that one exercise protocol for space travel be designed which would eliminate, or at least reduce, the physiological adaptations in both the musculoskeletal and the cardiovascular system (Thorton, 1990). To date, however, exercise has concentrated most on aerobic conditioning and has had limited success in addressing bone demineralization in space. This may be due largely to the absence of impact loading during any exercise program. To address this issue, Cavanagh et al. (1992) studied the efficacy of the three major exercise regimes currently used during space travel.

In this study, overground running, stationary rowing and stationary cycling were examined to determine the peak loads and loading rates under the feet. Force-time patterns were recorded for each activity and the peak load and loading rates for each exercise were determined (Figure 2.5). The average peak ground reaction force for running, cycling and rowing were 1,628 N, 371 N and 307 N, respectively. Mean values for loading rates achieved in the same three exercises were: 5,409 N/s for running, 1,558 N/s for cycling and 2,912 N/s for rowing. The results clearly show that running provides the highest loads and loading rates - magnitudes higher than that of either cycling or rowing. The authors hypothesize that these large forces applied to the body may be
critical for bone homeostasis and by choosing exercises with low magnitudes for the load and loading rate, the effectiveness of the exercise in combating bone demineralization in space is decreased.

![Force Profiles and Loading Rates for Running, Cycling and Rowing](adapted from Cavanagh et al., 1992)

**Figure 2.5:** Force Profiles and Loading Rates for Running, Cycling and Rowing (adapted from Cavanagh *et al.*, 1992)

Although the study by Cavanagh *et al.* (1992) clearly shows the advantages of running with respect to high impact force, the carryover of these results in a microgravity environment is unclear. Before running or any other high impact activity is implemented as an in-flight exercise, the ability to obtain 1G like response in space must be elucidated. As reported by Greenisen and Edgerton (1994), studies on the NASA KC-135 with instrumented treadmills shows forces of 95%, 120% and 150% of the subject's weight on Earth were found for walking and running during the periods of microgravity. These findings fall short of the expected 2-4 times body weight normally experienced during 1G running (Nigg, 1983). Better instrumentation and further studies are necessary to determine the validity of achieving high impact forces in space.
2.5 Exercise and Bone Mineral Density

The effect of the mechanical environment on bone mass has been extensively studied. One of the most powerful stimuli for bone remodeling is physical activity. When the physical demands are altered or the manner in which the bone is loaded changes significantly, bone will remodel in order to adapt to the changing functional environment (Brown et al., 1990). The threshold levels of activity associated with positive or negative changes in bone, however, are poorly defined due to the complexity of the loading history as well as interactions with age and the physical and physiological status of individuals (Forwood and Burr, 1993). Recent investigations have attempted to define a structure-function relationship between bone morphology and exercise (Gutin and Kasper, 1991; Smith and Gilligan, 1991; Snow-Harter and Marcus, 1991). These efforts all rely on the assumption that exercise increases bone mass which will in turn reduce bone fragility (Forwood and Burr, 1993; Bertram and Schwartz, 1991).

Many investigations support the hypothesis that exercise can effectively increase bone mass. In fact, numerous cross-sectional and longitudinal studies have been carried out in order to examine this relationship. Results, however, have been contradictory and in many case inconclusive. In order to devise more effective physical activity interventions for to maximize peak mass and reduce bone loss in post menopausal women, Drinkwater (1994) proposes several principles to which all protocols must adhere. According to Drinkwater, training must be specific to the site of interest. The bone of interest must also experience loads to which it is unaccustomed in order for the activity to be osteogenic. One factor which may account for the differing reports in the literature is the initial bone mineral density of the subjects. Inactive individuals have the capacity to achieve greater improvement in BMD and the benefits of the exercise will be overestimated. While the author concedes that there are no data available to verify these principles, sufficient data exist to support them.

2.5.1 Exercise and High Impact

Bone mineral density (BMD) has been found to be elevated in athletes who sustain high impact loads in training. Slemeda and Johnson (1993) showed an 8 to 14% greater
BMD in the trunk, legs and pelvis of figure skaters compared to control subjects. Gymnastic training has shown to significantly increase bone mineral density in the femoral neck and lumbar spine up to 8% (Nichols et al., 1994; Dyson et al., 1997). Studies also indicate that young female runners have similar lumber BMD to that of non-runners but have higher BMD in the distal femur and proximal tibia (Heinonen et al., 1993). Older, life-long runners, both male and female, have shown greater BMD in the lumbar spine (Michel et al., 1989). Walking above the anaerobic threshold level for postmenopausal women has similar effects (Hartori et al., 1993).

Heinonen et al. (1996) showed that exercises which load bones with a rapidly rising force profile improve skeletal integrity. High impact jumping and stepping exercises with forces ranging from 2.1 to 5.6 times body weight were performed three times a week for a period of 18 months with healthy, premenopausal women. After completion of the exercises, BMD increases in the lumbar spine, femoral neck, distal femur, patella, proximal tibia and the calcaneus ranged from 1.4 to 3.7%. While these increases were significant, they are not unlike values found for strength training and conflict with several other investigations which found training regimes to effectively increase muscular strength but not BMD (Blumenthal et al., 1991; Pruitt et al., 1995). The authors attribute the differences in results to the high loading rate type of exercise. Other differences could be due to the age, nutrition and hormonal status of the women participating in the study. Cross-sectional studies support the fact that high-impact training that produces high strain rates and forces to which the body is unaccustomed enhances bone formation (Heinonen et al., 1993). Additionally, longitudinal studies utilizing jumping exercises have shown increased trochanteric BMD as well as maintenance of the lumbar spine (Bassey and Ramsdale, 1994; Grove and Lunderee, 1992).

The particular type of exercise which is best suited to enhance BMD has been a subject of debate in the literature (Chilibeck et al., 1995). Endurance training is thought to be beneficial through repetitive impact loading. Studies in this area have focused on postmenopausal women and endurance-type, impact loading exercises. With varying amounts of exercise and duration, BMD and bone mineral content (BMC) in the lumbar
spine, as measured by dual photon absorptiometry, were found to increase or be maintained at the current level (Grove and Londeree, 1992; Krolner et al., 1983; Dalsky et al., 1988). Walking intervention studies have been less successful in confirming the role of exercise in increased BMD (Cavanaugh and Cann, 1988; Martin and Notelovitz, 1993). Positive gains in bone mass were only found in walking in which the anaerobic threshold was passed (Harori, 1993). The findings of this study strongly suggest that endurance training has a direct effect on bone remodeling. Several studies have demonstrated that repetitive impact exercises can be detrimental to the skeletal integrity. Male distance runners averaging over 50 miles a week were found to have lower BMD in their lumbar spine than non-runners (Bilanin et al., 1984). In women, hormonal levels also contribute to the damaging effects of endurance exercise (Grimston et al., 1993). Although there is no conclusive evidence, these and other studies seem to indicate that there may be a threshold level of exercise. It is hypothesized that within this threshold, exercise will have an osteogenic effect. Exceeding the threshold limit during exercise, however, will result in bone resorption and damaging effects to the body.

2.5.2 Strength Training Exercise

Proponents of impact-loading type training hypothesize that the most efficient means of promoting bone formation is a combination of gravity and an increase of body weight. Alternatively, strength training is thought to have the same effect through increased muscle pull. Cross sectional studies of athletic subjects have verified that strength training athletes have higher BMD than endurance trained athletes (Chilibeck et al., 1995). Strength training is thought to increase bone through local effects. Correlations have been found between hip adductor and quadriceps strength and hip BMD, back extensor strength and spine BMD and grip strength and radius BMD (Snow-Harter et al., 1990; Eickhoff et al., 1993; Sinaki and Offord, 1988). Although high impact training is looked more favorably upon as an intervention for decreased bone mineral density, Snow-Harter et al. (1994) demonstrated that both running and weight training
produce moderate increases in the lumbar spine. Additionally, a significant number of longitudinal studies have demonstrated that muscle can have direct effect on bone through tension at its attachment (Ryan et al., 1994; Pruitt et al., 1992).

Chilibeck et al. (1995) report that site specific strength training is effective for increasing the bone mineral density in the forearm whereas whole body training favors the lumbar region. Furthermore, the authors state that males show greater increases when placed on a strength training program compared to women and frequency and intensity are important factors affecting bone mass. Exercise should be performed at least three days a week at an intensity of at least 60% in order to maintain sufficient strain magnitudes to stimulate bone formation.

2.5.3 Exercise and the Calcaneus

The calcaneus is the largest of the tarsal bones and is a crucial link between the body and the ground and also serves as a lever for many extrinsic muscles of the foot. It is comprised of a thin cortical shell surrounding a core of trabecular bone. This orientation and structure enables the calcaneus to provide strength, shock absorption and stability to the foot and body. The calcaneus is loaded in all weight bearing activities and appears to be sensitive to inactivity and unloading. These facts indicate that this bone may be ideal for studying the effects of high impact loads on bone mineral density. Williams, et al. (1984) followed 7 males runners who maintained consistent running routines as they prepared for the Hawaii Marathon. Single photon absorptiometry was used to measure the bone mineral content (BMC) of the calcaneus before and after the training period. The average increase in BMC of the calcaneus for these runners was 3.15%. Dalen and Olsson (1974) completed a similar study on a group of cross-country runners using dual energy x-ray absorptiometry (DEXA). A mean difference of 21% was measured between calcaneal BMC of runners and controls.

Hutchinson et al. (1995) studied the influence of daily activity on calcaneal bone mineral density in terms of site specific loading. Fifty three healthy males participated in this study and were followed for a nine week period. Daily activities were assessed using digital stepmeters which were worn during all upright activities. Subjects were asked to
log all activity and describe any type of exercise they participated in. Subjects were classified into low-load and high-load groups according to the types of exercise they completed. Calcaneal bone mineral density was measured three times during the nine week period on the right foot and twice on the left. Calcaneal BMD was 12% higher in the high-load group compared to the low-load subjects. No significant correlations were found between BMD and exercise minutes in either group or total exercise minutes in the high-load group. Only the amount of time spent doing high impact exercise for subjects in the high-load group was slightly correlated to calcaneal bone mineral density (r=0.41). Results of this investigation appear to confirm Drinkwater's hypothesis concerning site specific activities, i.e., high calcaneal loading activities increase calcaneal bone mineral density.

2.6 Jumping Exercises

Jumping is a functional component of many activities. In particular, jumping plays a critical role in athletic events such as gymnastics, track and field, volleyball, basketball and aerobics. Jumping activities are composed of three phases: push-off, flight and landing. Execution of each of these stages is important for performance and can be influenced by a multitude of factors including technique, height, velocity, foot contact, surface stiffness and shoe type (Caset and Bates, 1995). Biomechanical analyses have been conducted to examine the characteristics of different jump types and specific techniques. Results of these studies have focused on ground reaction forces, lower extremity kinematics and kinetics and energetics.

Several different types of jumps have been investigated in the literature. The countermovement jump refers to a vertical jump which begins with a movement in the opposite direction (towards the ground). Squat jumps, on the other hand have no counter movements and begin in a squat position. Drop jumps are similar to countermovement jumps but begin from a distance above the floor. Of particular interest during jumping, and more specifically during drop jumps, is the landing during which the body experiences high impact forces which increase with velocity. The landing phase, if not executed properly, can be detrimental and result in injury to the performer (Dufek and Bates, 1991).
Although there is no definitive relationship between load, load rate and injury, the high incidence of musculoskeletal injuries in athletes participating in landing events cannot be ignored (McAuley et al., 1987).

2.6.1 Biomechanical Data Related to Jumping and Landing

Many researchers have attempted to gain insight into the mechanics and coordination of jumping by using a combination of force, kinematics, kinetics, EMG analysis and modeling. These investigations have focused on athletic performance and have been used as a means of identifying optimum technique. In a study by McNitt-Gray (1993), the landing techniques of gymnasts were examined by comparing kinetics of drop landings from three separate heights. The most notable finding of this investigation indicated that the peak extensor moments and amount of work done at the joints increased with impact velocity (height). This increase was found to coincide with the increase in joint flexion, peak angular velocities and peak ground reaction forces. A consistent temporal sequence of peak extensor moments and angular velocities was found suggesting that these variables are independent of impact velocity.

Bobbert et al. (1987a,b) have demonstrated the effects of landing technique and height on drop landings. They have defined two separate types of drop jumps: the bounce drop jump (BDJ) in which the performer attempts to reverse the downward velocity as soon as possible and the counter drop jump (CDJ) where a larger downward movement is executed. The longer push-off duration of the CDJ causes greater angles at the hips and knees and lowers the center of mass to a greater extent. For shorter push-off times, the average vertical acceleration of the center of mass is larger due to increased moments at the knee and ankle joints which are accompanied by large power outputs. This indicates that drop jumps, and in particular BDJs, are an effective training method to improve mechanical output at the knee and ankle joints; however, optimal drop height must be determined.

Kinematics of the landing phase have been described by McKinley and Pedotti (1991). They have identified a specific goal for landing: to land in a manner that allows
for maintenance of stability and minimization of stress on the body upon impact. Each individual may adapt unique strategies to accomplish this goal; however, a common strategy was found for joint motion. As the body prepares for contact with the ground, maximum extension of the leg is evident. At landing, the limbs were found to be slightly flexed.

2.6.2 Impact Force During Landing

Jumping can be used effectively to study the implications of high impact forces on the body. The force profile of landing from a jump is characterized by two peaks of high intensity and is largely dependent upon landing patterns. Two main techniques have been observed: toe-heel and flat foot. Toe-only and heel-only landings occur less frequently. Flat foot, toe-only and heel-only produce unimodal force curves while the typical toe-heel landings reveal dual-peaked curves (Dufek and Bates, 1991). The initial peak is associated with forefoot contact and the second peak with the hind foot (Dufek and Bates, 1990). An abundance of information can be found in the literature concerning the magnitude of force in different landing activities. Panzer (1987) reported 12.3 to 15.1 times body weight for impact peak in athletes performing back somersaults. Aerobic dance movements revealed single peak forces of 0.98 to 1.98 times body weight for low and high impact methods respectively (Ricard and Veatch, 1990). The repetitive nature of this activity can have serious consequences regarding injury.

Vertical impact forces during jumping can be attenuated by modifications in technique. Mizrahi and Susak (1982) emphasize the role of joint motion and muscle action in diminishing peak forces. Following the conclusions of Radin et al. (1972), impulsive loading results in bone remodeling which stiffens the trabecular bone and therefore decreases shock absorption. Adequate shock absorption must be obtained by means of bone, soft tissue and joint motion. According to their research, toe-heel landings achieve these goals and help to reduce joint degeneration and bone fracture.
2.6.3 Jumping in Microgravity

Little is known about jumping exercises in microgravity and further investigation is needed to determine whether the high impact forces delivered to the body during this activity can help combat space flight induced osteoporosis. To understand the mechanisms of jumping in microgravity, Vrijkotte (1991) conducted a study comparing jumping in simulated microgravity to jumping in 1G. Two different types of jumps were analyzed in this study - the squat jump and the countermovement jump. Peak loads, loading rates and tibial accelerations were measured in 1G and in two simulated microgravity conditions using different gravity replacement systems and pulling the subject downward towards the force plate with a force equivalent to 50% body weight (Table 2.3). Peak loads during landing were 55% less in simulated gravity when compared to 1G and countermovement jumps were higher than squat jumps in both conditions. Similar trends are observed for loading rate and tibial acceleration.

<table>
<thead>
<tr>
<th>Average Values</th>
<th>1G</th>
<th>Simulated Microgravity Condition 1</th>
<th>Simulated Microgravity Condition 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>CMJ</td>
<td>SJ</td>
<td>CMJ</td>
</tr>
<tr>
<td>Peak Force (N)</td>
<td>3,410</td>
<td>2,999</td>
<td>2,043</td>
</tr>
<tr>
<td>Loading Rate (kN/s)</td>
<td>157</td>
<td>114</td>
<td>103</td>
</tr>
<tr>
<td>Tibial Acceleration (g)</td>
<td>73.4</td>
<td>62.7</td>
<td>55.8</td>
</tr>
</tbody>
</table>

**Table 2.3:** Average Peak Loads, Loading Rates and Tibial Acceleration in 1G and Simulated Microgravity


An important finding of this investigation was that jumping caused higher peak forces and loading rates in simulated microgravity than running. This has several implications. First, if it is indeed true that high impact forces are directly related to strains
experienced by the bone, the higher the impact force, the more likely the bone strains will exceed the osteogenic level as prescribed by Rubin and Lanyon (1987). Therefore, jumping may be a more viable means of counteracting bone demineralization in space. Secondly, as the ability of the zero gravity locomotion system to truly simulate microgravity has not been assessed, the likelihood of forces of similar magnitudes being reached is unknown. Running has been shown to significantly increase bone mineral density in the lumbar spine and proximal tibia and femur (Michel et al., 1989; Heinonen et al., 1993). If impact forces cause this phenomenon, it can only be assumed that jumping has the capability to do the same. Also, if the microgravity environment has a measurable damping effect on the impact force decreasing the magnitudes during running, jumping may be the ideal countermeasure for bone demineralization.

2.7 Shock Attenuation

The effect of impact loading on the body has been previously discussed; however, the effect of the imposing forces on the skeleton requires further review. Contact with the ground produces impact forces which in turn cause transient shock waves to travel through the body from the feet to the head. Repetitive loading generates intermittent shock waves which are propagated through the musculoskeletal system and attenuated by the shock absorbers of the body. Under normal conditions, shock transients are critical for the regulation of bone structure. The axial skeleton is the primary structure responsible for the transmission of these shock waves while the soft tissue, joint motion, muscle action, and articular cartilage serve as shock absorbing systems. Although the absorptive properties of the individual tissues have been explored, the characteristics of the body as a whole is unclear. Several investigators have postulated that the body’s inability to transform the transient forces experienced during repetitive loading activities leads to low back pain, osteoarthritis and other degenerative joint problems (Light et al., 1980; Radin, 1972; Voloshin and Wosk, 1982).

Accelerometers mounted on the body have been used to assess the propagation and attenuation of force invading the musculoskeletal system by measuring bone vibration. Light et al. (1980) attached two low mass accelerometers to the tibia: one mounted on
the skin and the other directly coupled to the tibia through two Kirschner wires driven into the bone. Acceleration measurements were obtained while the subject walked with different footwear. Simultaneous accelerometry recording were taken at the head with a bar held between the teeth. Several important results are reported in this investigation. First, the observed transient depends greatly on the footwear although all tibial recordings ranged from 2-8 G. In harder heels, the data revealed substantial tibial shock waves which were greatly attenuated with a loss of high frequency components as they reach the level of the head. With respect to the ability of the skin mounted transducers to reliably predict the bone vibrations, the authors state that these recordings are “...broadly similar to those from the tibia, but show loss of high frequency components and some overswing. They suffice to give the order of magnitude of the transient, but not the rate of loading.” In a similar investigation, Lafortune et al. (1995) reported that skin mounted transducers estimated peak tibial shock 2.1 times greater than bone mounted transducers with a 5 millisecond delay. Additionally, the response was inconsistent across subjects. However, frequency transformations of the skin transducer signals matched those obtained from the bone. Valiant et al. (1987) suggested a magnification or amplification factor to correct for signal distortion in the skin mounted transducers. Using an accelerometer with extremely low mass can also help alleviate these problems.

The attenuation of shock waves through the human locomotor system has been evaluated by several investigators. Wosk and Voloshin (1981) evaluated the attenuation at the knee and the knee to forehead. These values were determined as the ratio between signals in the tibial tuberosity and the medial femoral condyle and the tibial tuberosity and the forehead, respectively. Values reported were the peak to peak acceleration signals during the first cycles. Knee attenuation ranged from 1.24 to 1.56 while attenuation from the knee to forehead was significantly greater, 2.53 to 3.74. The ratio of shock peaks from the shank to the head are found to decrease for more severe impacts from faster running and greater stride lengths (Bhattacharya et al., 1980; Hamill et al., 1995). Accelerometric techniques can also be used to quantify the effectiveness of shoe inserts in reducing shock transmission (Voloshin, 1981). Frequency analysis and transfer functions
have been employed to examine shock transmissibility during running (Lafortune et al., 1995b). The relationship between the tibial acceleration and ground reaction force was confirmed in five physically active subjects at the individual and group level. Transfer functions indicated that frequencies higher than 30 Hz were transmitted to the tibia during running and those higher than 60 Hz were damped.

Although the ability of the body to control shock transmission relies on the unique shock absorbing properties of the heel pad and other viscoelastic structures, it is also reliant upon body posture. Running with flexed knees has shown to decrease stiffness in the lower limbs thus diminishing the transmission of shock from the feet to the head (McMahon et al., 1987). Lafortune et al. (1996) studied the effect of knee angle at heel strike on the shock transmission characteristics of the body. Using a human pendulum device (Lafortune and Lake, 1995), impacts of varied severity were delivered to the foot with the knee extended as well as flexed at 40 and 60 degrees. Larger knee flexion angles at impact improved the shock absorption demonstrated by a reduced head to shank ratio of peak accelerations, reduction of the effective axial stiffness of the body and improved filtering of all frequencies above 5 Hz. The gain/attenuation profiles revealed that the higher frequency components of the shock were damped to a greater degree than the low frequencies.

The role of the ankle in shock absorption has been studied during landing from a vertical jump by measuring acceleration on the medial calcaneus and the distal tibia (Gross and Nelson, 1988). Shock waves were measured during two different landing styles: forefoot only and forefoot followed by heel contact. Although ankle kinematics were similar for both landings, force and acceleration measurements were remarkably different with non-heel landers experiencing only one peak transient. The results suggest that landing without heel strike is an effective means of reducing shock transients experienced by the body.

2.8 Zero Gravity Simulation

The microgravity experienced in space cannot be duplicated on Earth - only approximated. Despite this fact, a great deal of insight has been gained regarding the
response of the body to altered gravity states through the use of numerous ground-based zero gravity simulations. Additionally, as in-flight experiments become more and more cost prohibitive, ground-based studies are more appealing to researchers wishing to prove their hypothesis regarding the physiological adaptations in space. Simulators are not only important for the study of the body's response to space but also for training of the astronauts and the assessment of human performance in the altered environment of space. Extensive studies have been conducted using various means of simulating zero gravity with clinical findings similar to those measured after both short and long periods of time in space. These include: bed rest, immobilization, free fall and partial suspension systems. A brief description of these methods along with their advantages and disadvantages will be reviewed.

2.8.1 Bed Rest and Immobilization

Bed rest is probably the most widely used form of zero gravity simulation. Currently, more than 190 bed rest studies have been conducted with more than 2,200 subjects ranging in age from 18 to 65 (Nicogossian, 1994). Although gravity is still present, these studies have shown remarkably similar results to space flight. These include muscle atrophy, bone demineralization, redistribution of fluids and body mass, decreases in red blood cell mass and orthostatic intolerance upon returning to an upright position (Sandler and Verikos, 1986). Various bed rest studies are performed in the head-down tilt position. Compared to the horizontal position, head-down tilt tests result in a greater and more rapid cardiac deconditioning and are more applicable to some of the changes which occur earlier in flight (Kakurin et al., 1976; Gazenko and Grigoriev, 1980). Bed rest studies followed by centrifuge of up to +3G are also used to assess the response of large accelerations after unloading to simulate Space Shuttle take-off and landing.

Immobilization techniques have been useful in determining the effects of hypokinesia. Animal studies have been performed utilizing partial or full body casts. Dickey et al. (1979) immobilized primates in horizontal casts for two to four weeks with resultant adaptations similar to those found in bed rest studies. Confinement of animals to cages, similar to immobilization, has also been used to examine the effects of hypokinesia.
on the musculoskeletal system. Rats confined in their cages for up to four months demonstrate weight loss, decreased size of the gastrocnemius and decreased protein content in muscle (Fedorov and Shurova, 1973). Dogs confined for six months show significant skeletal abnormalities — especially in the femur and tibia (Novikov and Vlasov, 1976). The suspension technique is also utilized as a means of immobilizing animals. Suspension is commonly used in rat models where the animal is suspended by the tail so the rear limbs are completely non-weight bearing and unrestricted (Morey, 1979). This method is generally preferred to immobilization or confinement because it provides a more controlled experimental environment.

In humans, the effect of immobilization is clearly illustrated in patients with paralysis, fracture or other surgical procedures which require non-weight bearing activity. An analysis of bone mineral density of the calcaneus after four weeks of non-weight bearing of the lower extremities showed an average loss of 7.2% (Ozaki et al., 1994). Following six weeks in a body cast, patients lost 1-2% of total body calcium as measure by urinary excretion (Detrick et al., 1948). Although osteopenia has been documented in some patients, other reports indicate no significant decrease in bone mass after several months of immobilization (Vico et al., 1987). These contradicting reports emphasize the fact that more efficient and accurate means of simulating microgravity on Earth must be developed to reliably measure the human response to weightlessness.

2.8.2 Free Fall

Free fall devices are capable of producing momentary periods of zero gravity on Earth providing the fall is of long enough duration. The first to experiment with this idea was von Diringshofen who accelerated an aircraft towards earth from a high altitude (Sharpe, 1969). This produced eight seconds of microgravity. Using this concept, Heinz and Fritz Harber later developed the parabolic flight trajectory method which is commonly used in NASA KC-135 flight experiments to conduct studies in zero gravity (Davis and Cavanagh, 1993). Occupants of the aircraft experience approximately thirty seconds of
microgravity at the apex of the trajectory but must endure high G levels as the plane dives and begins the next trajectory. This can result in motion sickness for the individuals on the aircraft.

Vertical motion devices which allow for free fall are alternate methods simulating microgravity. These include drop towers, sub-gravity towers and elevators. Both the drop tower and the sub-gravity tower work on the concept that by suspending a payload from a high height and releasing it, brief periods of microgravity are created. These devices are suitable for studying the effects of weightlessness on the instrumentation carried aboard the spacecraft and generally not employed for human studies. Ascending and descending elevators, however, can be employed for brief physiologic tests.

2.8.3 Suspension Systems

Numerous forms of cable suspension systems have been developed in which the subject is suspended from a scaffold either in the upright or horizontal position. Vertical suspension systems involve partially unweighting the body as the subject performs a given activity. This has been accomplished by several means. Camacho et al. (1969) used parachute harness and a single cable attached overhead (Figure 2.6a). In a simplified device, He et al. (1991) used a series of spring attached to a bicycle saddle over which the subject straddled their legs. The springs, when tensioned, were capable of unloading the body to varying degrees simulating reduced gravity states (Figure 2.6b). In essence, these simulators rely on the fact that by counterweighting the body, the normal force and traction on the ground are decreased (Davis, 1991). The body, however, still experiences the weight of 1G force. A further drawback is the limitation of movement in the sagittal plane due to the interference of the supporting cables.

To further negate the effects of gravity and increase the freedom of movement, horizontal cable suspension systems have been developed. The first of this type of suspension system was the inclined lunar sidewalk (Figure 2.7). In this device, the subject walks in a parallel to the ground suspended from a curved bar and cables attached to an overhead trolley (Hewes, 1969). Although the efficacy of this suspension system has been questioned, the inclined lunar sidewalk was extensively used to simulate moon walks and
Figure 2.6: Vertical Suspension Zero Gravity Simulators.
investigate varying gravity levels in centrifuge (Davis and Cavanagh, 1993). Alternate
zero gravity simulators place the subject in a supine position attached with some means of
counteracting the weight of the body (Figure 2.8). Initial supine systems suspended the
body using counterweights. Performance in this system, however, was difficult especially
in terms of energy consumption due to the up and down acceleration of the weights. To
overcome this problem, Grigor'yev et al. (1987) used elastic cords attached to several
segments of the body to unweight them. Cavanagh et al. (1989) used a similar approach
but with separate cord for each individual segment thus eliminating the interdependence
between adjacent segments for motion.
2.9 Summary

It is evident from the literature that has been review in this section that the problem of bone loss in space is one which cannot be ignored. Definite alterations in the bone structure and metabolism occur in a microgravity environment. If missions are to be taken to Mars and beyond (an issue which is being widely considered today), prevention of the physiological adaptations in space must be dealt with - especially if astronauts are to function normally upon their return to the Earth’s gravity.
CHAPTER 3

METHODOLOGY

The underlying objective of this research is to identify the effectiveness of jumping exercise protocols in the maintenance of skeletal integrity. In particular, the response of the lower extremity, the part of the musculoskeletal system which experiences the greatest loss of strength and function during long term space missions, to high impact jumping exercises will be examined in 1G and simulated microgravity. While other investigations have focused on the biomechanics of walking and running in reduced gravity (Davis, 1991; Cavanagh et al., 1992; McCrory, 1997), the current research proposes that jumping exercises may provide the necessary levels of strain, a known osteogenic stimulus, to prevent space flight induced osteoporosis. As previously noted, jumping activities impart high impact forces to the body. However, the effect of these forces on the internal bone strain is unclear. Therefore, the current research will, for the first time, document the relationship between externally applied loads and internal bone strains.

In order to examine the ability of jumping exercises to diminish bone demineralization in space, a multifaceted experimental approach is necessary. A means of simulating reduced gravity in a 1G environment must be developed. As discussed in Section 2.8, several zero gravity simulator designs exist. For the current investigation, a horizontal suspension system was chosen. The underlying concept of this device is that the effects of gravity on the body segments are negated by the opposing forces in the suspension cords. In the supine position, gravity is no longer acting on the long axis of the body and any measurements recorded in this direction are without its influence. The design of the zero gravity simulator should meet certain criteria. First, the proposed
simulator should mimic the conditions encountered during exercise in space in order to assess the effects of jumping in space. This device must also allow for high impact forces to be created under the feet during jumping. It is therefore necessary to create a means of tethering the subjects to the wall and producing artificial gravity. With the tethering system, the body will be accelerated back towards the supporting surface during the flight phase of the jump allowing for a high impact landing. Tethering systems are also used during exercise on the current space missions.

Once the device for simulating reduced gravity is designed, the measurement of relevant parameters must be considered. New techniques for quantifying internal bone deformation must be established. Although it would be desirable to mount a strain gauge directly on the calcaneus, this is prohibited by the shape and location of the bone. However, it is anticipated that an external bone extensometer will adequately measure calcaneal strains. A method of directly linking bone strain information with the external load requires the use of not only bone extensometers but also force plate measurements. Additionally, the transmission of the shock transients during landing will also be explored by placing accelerometers on both the tibia and the calcaneus. A complete description of the instrumentation and experimental methods follows.

3.1 Overview of the Zero Gravity Simulator

A zero gravity simulator (ZGS) was constructed using latex cord, rope, harnesses and braces to suspend subjects from a twelve foot high ceiling in a supine position (Figure 3.1). The ZGS consists of two distinct systems exerting forces on the subject. The suspension system, consisting of latex cords and ropes, supports the individual and negates the effect of gravity on the lower extremities. The gravity replacement system (GRS) tethers the subject to a force plate mounted on the wall. With the GRS, varied levels of gravity can be created in the ZGS. Each of these systems are calibrated according to the subject's anthropometric dimensions and the desired gravity level.
3.1.1 The Suspension System

The role of the suspension system is to unweight the subject. It was therefore necessary to devise a method of supporting subjects of varied height and weight with a constant upward force during jumping on a wall mounted force plate. Ceiling brackets were secured to a twelve foot high ceiling. A slot running along the length of the bracket allowed metal supports, to which the suspension cords were attached, to be adjusted to the appropriate position and then firmly clamped in place. Based on the work of Davis (1991), solid latex cords were chosen as the suspension material because of their elastic nature. The latex acts as a Hookean material so the deformation is proportional to the force. This property makes the latex an ideal material to meet the design constraints of adjustability and constant force.

\[\text{Figure 3.1: A Schematic Representation of the Zero Gravity Simulator Device Used to Study Jumping Exercises in Simulated Microgravity}\]
In order to determine the length of unstretched cord used in the ZGS, it was essential to calibrate the latex cords, set limits on the weight which would be supported by them and determine the final length of the cord based on the weight it supported (Appendix A). Initial attempts at using single strands of latex cord failed because of the high elasticity of the material and double stranded lengths were used in the final design.

The underlying assumption for the suspension system was that the force in the latex cord will not fluctuate during any activity in the zero gravity simulator and the system was designed to eliminate any unnecessary deviations from a constant force. Previous studies in a similar gravity simulator (Davis, 1991; McCrory, 1997) have analyzed locomotion. During walking and running, the limbs of the supine subject in the zero gravity simulator move in two directions: towards the ceiling and away from the wall. During jumping, the limbs only move in one direction - away from the supporting surface. The only time the center of mass changes its vertical location is during the countermovement when the flexion in the knees may cause the limb to be elevated. This however, has very little effect on the force being exerted on the body segment by the latex cords.

The motion of the center of gravity of both the thigh and calf segments in the vertical direction was determined from the location of the hip, knee and ankle markers for one subject during jumping (Figure 3.2). When the subject was "standing" erect in the zero gravity simulator, the cords were calibrated to match the weight of the lower extremity segments. The deviation from this calibrated "resting" position occurred when the subjects flex their knees causing the center of gravity to rise above the resting position decreasing the load carried by the latex cords. From Figure 3.2 it is evident that the greatest deviation in COG position occurred during the countermovement phase of the jump although there were deviations present during landing. The average deviation from the calibrated resting position (the point where the subject was standing straight on the platform) was approximately 8.2 cm higher and 3.2 cm lower for the thigh and 15.4 cm higher and 4.1 cm lower for the shank in the subject tested. This represents a maximum deviation in the cord length of 8.6 and 10.7% for the thigh and shank respectively.
3.1.2 The Gravity Replacement System

The second set of forces acting on the subject in the zero gravity simulator is due to the gravity replacement system (GRS). Similar to astronauts in a weightless environment, the suspended subject would not be capable of eliciting high impact forces on the ground (or in the case of the ZGS, the wall) due to the lack of forces pulling the individual towards the supporting surface. The GRS therefore, provides the necessary force for contact through a series of ropes and springs which can be tensioned to impart several levels of "artificial gravity." The ideal situation in the GRS is to provide a constant force equal to gravity on Earth. A more realistic approach would be to enable the subject to attain impact forces equivalent to those found during activity in 1G. This made it necessary for the gravity replacement system to provide acceleration during the flight phase of jumping to reverse the direction of the jump and allow the subject to land on the supporting surface. This approach is attainable if the activity of jumping is considered. As the subject jumps away from the wall, the springs are stretched and consequently exert a
greater force in the opposite direction of the jump. Once the force in the springs exceeds the force of the body accelerating outward, the subject will be pulled back toward the wall, contacting it with a certain velocity.

The GRS consists of two heavy duty springs with rope attached on either side. Rope is attached at the subject’s waist in front and behind, similar to the harness system used by astronauts in space. From the waist the rope goes through a pulley and is attached to the spring. A second rope, tied to the opposite end of the spring, travels to a pulley mounted in the ceiling brackets, back down to the supporting wall and is clamped in place by a mechanical locking device which hold the rope in place (Figure 3.2). Prior to being placed in the ZGS, the springs were calibrated by hanging a series of weights on the end and measuring the resulting displacement. A regression was performed yielding the following relationships:

\[
\text{Front Spring: } F_1(N) = 26.52 \times \text{(displacement)} + 148.13 \quad (3.1)
\]

\[
\text{Back Spring: } F_2(N) = 27.48 \times \text{(displacement)} + 135.23 \quad (3.2)
\]

where the displacement (in centimeters) is the difference between the stretched length and the resting length.

To tension the spring while the subject is suspended in the ZGS, the rope is pulled through the locking device while the highest end of the spring is stretched away from the suspended subject to the appropriate length - dependent on the subject's body weight and the desired gravity level. The length of each spring which would exert the appropriate force for each gravity level was determined in the following manner. The angle between the long axis of the body and the GRS ropes attached anteriorly (\(\alpha\)) and posteriorly (\(\beta\)) at the waist was measured using a goniometer (Figure 3.3).
Figure 3.3: Forces in Gravity Replacement System

The force in the GRS ropes due to the springs was resolved along the long axis of the body and set equal to the desired percentage of the subject's body weight (Equation 3.3).

\[ F_1 \cos(\alpha) + F_2 \cos(\beta) = F_R \]  \hspace{1cm} (3.3)

where
- \( F_R \) is percent body weight
- \( F_1 \) is the force in Spring 1 according to Equation 3.1
- \( F_2 \) is the force in Spring 2 according to Equation 3.2
- \( \alpha \) is the anterior angle
- \( \beta \) is the posterior angle
To balance the tethered subject in the zero gravity simulator, the sum of the vertical components of the force are equivalent (Equation 3.4).

\[ F_1 \sin(\alpha) - F_2 \sin(\beta) = 0 \]  

(3.4)

Using Equations 3.3 and 3.4 and substituting for \( F_1 \) and \( F_2 \), the length of the front spring \( L_{\text{front}} \) and the back spring \( L_{\text{back}} \) are calculated:

\[ L_{\text{front}} = \frac{F \sin(\beta)}{\sin(\alpha) + \cos(\beta)} + 5.58 + L_{\text{rest}} \]  

(3.5)

\[ L_{\text{back}} = \frac{F \sin(\beta)}{\sin(\beta \cos(\alpha) + \cos(\beta))} + 4.92 + L_{\text{rest}} \]  

(3.6)

where \( L_{\text{rest}} \) is the resting length of the spring (63.5 cm).

### 3.1.3 Wall Mounted Force Plate and Platform

A Bertec force plate (Bertec Corporation, Worthington, OH) was mounted in the wall below the suspension system. The "wall" was actually the side of an elevated walkway. The side of this walkway would not provide a sound surface upon which the force plate could be mounted. Therefore, a section was partitioned off and filled with cement. The cement extended out past the side of the wall creating a rigid platform. A steel plate was secured on the cement and leveled. The force plate was mounted on the steel and screwed into place. A wooden platform was designed to fit around the force plate to provide an additional surface upon which the subjects could land.

### 3.2 Pilot Studies

A preliminary experiment was performed for several reasons. Since this study required the construction of a zero gravity simulator, it was necessary to verify its accuracy, assess the instrumentation of the system and determine its ability to mimic a microgravity environment. Pilot studies were also carried out to determine the viability of jumping in simulated microgravity.
Six subjects (2 male, 4 female, average mass, 63.47 ± 12.02 kg) were outfitted with the leg supports and harnesses and suspended in the zero gravity simulator. The springs in the GRS were tensioned at perceived low, medium and high levels and the subjects were asked to perform a countermovement jump with both feet contacting on the force plate upon landing. Subjects were instructed to limit their knee flexion on the countermovement but were given no particular instructions on how to land other than to land "naturally." Ground reaction forces were collected from the wall-mounted force plate at 2400 Hz for the entire jump (countermovement and landing). A standing and a zero calibration file were collected prior to each jump. The zero ground reaction force data were averaged over a period of 0.1 seconds and this baseline was subtracted from the trial data. The standing file was used to calculate the "equilibrium" weight - the amount of tension in the springs pulling the standing subject toward the wall, expressed as a percentage of the overall body weight.

Spring tensions ranged between 30 to 100% of the subject’s body weight resulting in peak ground reaction forces, $F_{\text{max}}$, during landing between 1165 and 3106 N (Figure 3.4a) representing between 1.5 and 4.7 percent of the subject's body weight in 1G. The loading rate for each jumping trial at the various tension levels was calculated as the rate of change of force between the onset of landing and $F_{\text{max}}$ (Figure 3.4b). Several factors are apparent from these findings. Prior to this initial testing, the amount of tension in the springs the subject could comfortably endure during jumping exercises in the ZGS was in question. In previous studies (Davis, 1991), the maximum amount of tension allowable was 60% of body weight. In the current device with only two springs (as opposed to four in the previously mentioned experiment), springs could be tensioned to 100% body weight during jumping. This allowed for a wide spectrum of tension levels in the ZGS to be tested. Additionally, the subjects in the pilot study landed with both feet on the force plate. Upon further consideration, however, it was decided that only one foot (the right foot) should contact the force plate upon landing. The reasons for this are twofold. One of the underlying goals of this research was to determine the relationship between the external ground reaction forces and the internal bone strains. Since only the right leg would be instrumented during the experimental trials, it would then be necessary to
assume the force was equally distributed between the left and right side. By adjusting the position in the zero gravity simulator so only the right foot contacts the force plate and the left foot touches down on the adjacent platform, the exact force transmitted to the instrumented leg is known. Furthermore, comparison to the cadaveric drop test studies which measured force and strain variables on only one limb is more straightforward.

![Graph](image)

**Figure 3.4:** (a) Peak Ground Reaction Forces, Fmax and (b) Loading Rates, dF/dt in the Zero Gravity Simulator During Countermovement Jumps
One of the more interesting findings from these pilot data concerns the landing mechanics of the countermovement jumps. Before testing, it was assumed that all landings would occur in a "toe-heel" fashion - initial contact on the force plate would be made with the ball of the foot followed by the heel until the foot remained flat on the supporting surface. However, many subjects never contacted the surface with their heels, landing exclusively on their toes. This was attributed largely to the knee flexion angle upon landing and was explained by the bent knee posture caused by the simulator. More surprising was the fact that even on the floor, the same subjects did not contact their heels on landing. Thus, future jumping experiments in the zero gravity simulator and on the ground required careful instructions to the subjects regarding their landing. It was also noted that landing which occurred in a more "flat-footed" manner yielded higher loads and loading rates. For example, Subject 6, whose landing was nearly flat, achieved the highest peak loads and loading rates. Additionally, even though Subject 6 had the lowest mass, these observation held true whether the absolute value or the normalized value of these variables were examined. It was expected, therefore, that by utilizing different landing protocols, a wider response would be attained. This allowed for a greater spectrum of loads and loading rate which may be critical in determining the level of load necessary to exceed the osteogenic strain threshold.

3.3 Experimental Procedure

The underlying emphasis of this investigation was to determine the relationship between the external ground reaction forces. It was hypothesized that this relationship will be useful in ascertaining the types of activities which will help reduce space flight induced osteoporosis. Additionally, the effect of ZGS tension levels on the ground reaction force and other measured variables was assessed as well as the differences between simulated microgravity and 1G. The latter is an important factor if the data are to be applied to astronauts in space. As previously noted by Cavanagh et al. (1992), it is unclear whether or not the forces obtained in 1G locomotion can be obtained in a microgravity environment. The comparison between ZGS and 1G data will also give insight into the optimal tension level in the simulator.
The current experiment is multifaceted with several factors being investigated within the same trial. All twelve subjects performed the same jumping exercises in the zero gravity simulator and in 1G. However, four of the subjects were instrumented with calcaneal extensometers to measure calcaneal strain during jumping. These subjects executed the same jumping protocol as the remaining eight subjects, but the additional information related to strain was collected during the trials.

3.3.1 Statistical Analysis

3.3.1.1 Experimental Design

Experimental conditions consisted of the five spring tensions. In the zero gravity simulator these tension were: 45, 60, 75 and 100% of the subject's body weight. The fifth "tension" was the 1G trials. Three different types of landing were performed during the jumping exercises: (a) toe-heel landing on both feet; (b) toe-heel landing on one foot; and (c) flat-footed landings on two feet. Each jump was repeated three times in each experimental condition resulting in 9 jumps per condition with a total of 36 jumps per subject. Tension levels were randomized under the constraint that ZGS trials be randomly administered in the first part of the testing session followed by the 1G condition in the latter part (or visa versa). Additionally, all jumps at each tension level were completed before moving to the next tension. This constraint was imposed to prevent fatigue of the subjects from constantly changing the tension level and to shorten the experiment time by eliminating the necessity of suspending the subject more than once (a process which takes about 45 minutes). Landing types were randomized within each tension level (Table 3.1).

<table>
<thead>
<tr>
<th>Condition</th>
<th>Landing Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>ZGS 60%</td>
<td>T O T F O T F F O</td>
</tr>
<tr>
<td>ZGS 45%</td>
<td>F O T T O O F T F</td>
</tr>
<tr>
<td>ZGS 100%</td>
<td>O F O O F T F T T</td>
</tr>
<tr>
<td>ZGS 75%</td>
<td>O T T T F F O O F</td>
</tr>
<tr>
<td>1G</td>
<td>F T T O F O F T</td>
</tr>
</tbody>
</table>

Table 3.1: Typical Randomized Order of Experimental Conditions
The experimental design was a 2-factor repeated measures design with compound symmetry. The twelve subjects served as the random block factors with the fixed factors of landing type and spring tension. Each subject was exposed to all five tension levels with all three landing types repeated three times at each tension. Because each jump was repeated by every subject for each condition, there was variability within the subject as well as between subjects. Two variance components were therefore used to adjust for the correlation between responses obtained from the same subject. The mixed statistical model for analyzing the data collected in the zero gravity simulator and 1G is written as:

\[ y_{ijkl} = X_{ijk} \beta_{jk} + Z_{ijk} \gamma_i + \varepsilon_{i} \]  

This model is expressed with all variables in matrix form where \( y \) represents the univariate data, \( X \) is a known model matrix, \( \beta \) is an unknown vector of fixed-effect parameters, \( Z \) is the design matrix, \( \gamma \) is the vector of unknown random-effect parameters and \( \varepsilon \) is the unknown random error vector with a mean of 0 and variance \( \sigma^2 \). Subscripts represent the subjects (i), landing type (j), spring tension (k) and repetition (l).

### 3.3.1.2 Analysis of Variance

An analysis of variance (ANOVA) was performed on the following variables: maximum and minimum calcaneal strain, strain rate, peak calcaneal and tibial accelerations, peak force, loading rate, jump height, impact velocity and flight time. The alpha value used to determine statistical significance between treatment means was set at 0.05 for each variable. When data was not significantly different between conditions, paired t-tests were performed to assess the similarity in the variable (Appendix B).

### 3.3.2 Subject Selection

A power analysis was performed to determine the number of subjects in the study. The relationship between the sample size and the minimum detectable difference (MDD) for the maximum force, \( F_{\text{max}} \), and the maximum loading rate, \( dF/dt_{\text{max}} \), was established using the nQuery Advisor software (Statistical Solutions, Boston, MA). The largest standard deviation of the differences of least squares means was used to estimate the
MDD with a power of 80% (Figure 3.5). These graphs demonstrate that as the sample size increases, MDD decreases asymptotically. It is therefore apparent that there is less to be gained from recruiting more subjects - with eight subjects appearing to be acceptable. For the ZGS experiments, twelve subjects were chosen which enables a MDD of approximately 200 N for \( F_{\text{max}} \) and 25 kN/sec for \( \frac{dF}{dt_{\text{max}}} \) to be measured. Therefore, if the measured values of both these variables exceed the prescribed MDD, it can be concluded that there is a significant difference between trials.

![Graph (a)](image1.png)

![Graph (b)](image2.png)

**Figure 3.5:** Sample Sizes for Measuring Differences in Peak Load (\( F_{\text{max}} \)) and Loading Rate (\( \frac{dF}{dt_{\text{max}}} \)) (Based on data from Davis, 1991)

The twelve subjects were chosen based on age, weight and height. In order to adequately represent typical astronauts, it was necessary that the anthropometric data of all subjects fell within the range of those values measured for astronauts (Table 3.2a). Six females and six males who met this criteria were selected (Table 3.2b). All subjects were in good health and physically fit. Four of the twelve subjects participated in the invasive bone strain experiments in addition to the simulated zero gravity jumping exercise protocol. Prior to participating in the study, subjects completed a medical questionnaire (Appendix C), read and signed the Informed Consent form (Appendix D) and had the procedures fully explained to them and any questions answered.
### ASTRONAUT DATA

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<th>Height (cm)</th>
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<th>Weight (kg)</th>
<th>Height (cm)</th>
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### SUBJECT DATA

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<th>Height (cm)</th>
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**Table 3.2**: Age, Mass and Height Statistics of: (a) Astronauts and (b) Subjects

The right leg of the subject was prepared for the experimental trials. EMG electrodes were placed on the medial gastrocnemius and the vastus lateralis. Because these electrodes would be covered by the leg braces which may have disrupted their placement or caused pain to the subject, they were covered with a foam "donut" which was secured with pre-wrap and tape. Accelerometers were taped on the distal third of the
tibia and on a bony portion of the lateral calcaneus. These transducers were also fastened with tape and pre-wrap. The foot was wrapped with cloth adhesive tape. The step was taken because the sweat from the foot tended to disrupt the placement of the instrumentation and the tape eliminated this problem. Retroreflective markers were positioned on the greater trochanter, femoral epicondyle, lateral malleolus, head of the second metatarsal and the posterior heel using two-sided tape.

### 3.3.3 Anthropometry

In order to correctly attach the suspension cords correctly, several anthropometric measurements were necessary (Appendix E). After recording the height and weight of each subject, the distance from the floor to the center of gravity of the calf and thigh were determined. The centers of gravity were measured according to Clauser et al. (1969) as 37.2% from the hip and 37.1% from the knee for the thigh and calf segments, respectively. Additionally, the position of the waist harness and chest harness on the subject were ascertained. This allowed for the suspension cords and ropes to be positioned directly over each subject completely balancing the subject in the zero gravity simulator. If the cord was not attached at the center of gravity of the lower extremity segments, a resultant couple would cause the limb to rotate thus reducing the effectiveness of simulating reduced gravity. The leg braces were also designed so the attachment of the latex cord to the lower extremity segments would fall approximately at the center of gravity on all subjects.

In order to accurately negate the effect of gravity through force in the latex cord, the weight of each segment was determined from linear regression equations (Vaughan et al., 1992). Based on the mass of the subject and assuming uniform density, several linear dimensions were used to predict the calf and thigh mass. These included: thigh length, midthigh circumference, calf length, calf circumference, knee diameter, foot length, malleolus height and width, and foot breadth. The right and left legs were assumed to be symmetrical and measurements were recorded only from the right side.
3.3.4 In Vivo Strain Gauge Preparation

The four subjects who were instrumented with the calcaneal strain gauge underwent a surgical procedure to implant two K-wires, upon which the transducer (See Section 3.4.3) would be mounted, in the bone. Two hours prior to the surgery, prophylactic antibiotics were administered to decrease the chance of infection at the wound site. A licensed orthopaedic surgeon performed the procedure at the Cleveland Clinic Foundation. To ensure a sterile environment, the lateral heel was draped and scrubbed prior to the implantation and a local anesthetic was injected in the peroneal nerve. The lateral heel was viewed on a C-arm fluoroscope to determine the orientation of the calcaneal trabeculae. Since the trabeculae are oriented in the direction of the principle force transmitted through the bone, it was decided that the K-wires should be aligned with the trabeculae to acquire the most effective measurement of strain. A custom made jig was used to position the pins at a predetermined distance of 0.524 inches apart (Figure 3.6).

![Custom Made Drilling Jig and Pin Implantation](image)

**Figure 3.6:** Custom Made Drilling Jig and Pin Implantation

Once the area was completely numb, the two K-wires were drilled into the bone. Using the drilling jig and a surgical drill, each wire was drilled approximately half way
through the bone. This was verified by a fluoroscopic image (Figure 3.7). Once the wires were in the correct position, the jig was removed and the insertion site was covered with an antibiotic strip to prevent infection. Subjects were able to ambulate freely immediately after the surgery and were in no pain. Caution was taken during the jumping experiments to assure that the pins were not disrupted and the subject was in no pain. Upon completion of the protocol, the pins were removed by the doctor, the wound was cleaned and dressed. The subject was given instructions on wound care and told to finish their antibiotic dose.

![Figure 3.7: Fluoroscopic Image of K-Wire Placement in the Calcaneus](image)

### 3.3.5 Subject Suspension and GRS Tensioning

Once the body segment masses were determined and all instrumentation attached to the right leg, the subject could be suspended in the zero gravity simulator. Subjects lay supine on a portable bed placed under the suspension cords. I-bolts in the ceiling brackets were positioned over the center of gravity of the thighs, calves and at the point of attachment of the chest and waist harness. Specially designed limb braces were secured on the right and left thigh and calf. These braces were constructed in a manner to allow the latex cords to act directly at the center of mass of each body segment. Using a Chatillion® force transducer, the latex cords were stretched to equal the weight of each
segment. The cord was then attached to the brace using a carabiner and an adjustable knot. The waist and chest were fitted with a climbing harness and a helmet was worn on the head. Ropes were attached to these supports in order to stabilize the upper body and the torso. Wooden planks were placed under the waist and chest harness to distribute the forces due to the pull of the springs and increase comfort while suspended and tethered to the wall. The bed was lowered and the subject hung in the zero gravity simulator. Vertical adjustments were made in the latex cords to assure that the subject's right foot was centered on the force plate and the left foot contacted only the platform adjacent to the force plate. Alignment of the head, torso and legs were checked and any further refinements made with the mountain climbing ascenders attached to the waist and chest harnesses. A rope was attached to a weighted electromagnet behind each subject at the waist to limit the countermovement to 15% of the body weight. All measures were taken to make the subject comfortable. Figure 3.8 shows a subject suspended in the zero gravity simulator with all instrumentation attached.

The instrumentation was connected to the subjects so it was out of their way and would not hinder their performance. The subjects were asked to stand with their knees straight so the springs could be tensioned to the appropriate level. The springs were pulled away from the subject and the length measured. When each spring was at the correct length, the subjects were instructed to relax by flexing their knees. Subjects were also told that they could relieve the pressure from the tension in the springs at any point during the trials by bending their knees and dangling their feet in the air.

3.3.6 ZGS and 1G Trials

Once subjects were acclimated to the zero gravity simulator, several practice jumps were performed. The subjects was instructed on what they were doing improperly. Practice jumps continued until the subjects could land properly for all three types of landings. Most important was the fact that the subjects contacted the force plate with their heels on every landing. Before each jump, a zero reading was captured. During this time the subjects were told to remove their foot from the force plate either by placing both feet on the platform or flexing their knees causing their feet to dangle in the air. Subjects
with the calcaneal strain gauge used the latter approach. An equilibrium value was recorded while the subject was standing with their full "weight" on the force plate. These data were used to confirm the force pulling the subject towards the wall matched the expected value as determined by the length of the springs. They would also supply a zero reading for the accelerometer data.

Figure 3.8: Subject Suspended in the Zero Gravity Simulator

Jumping exercises occurred in a sequential manner. Subjects performed nine jumps at each tension level - three of each type landing. Subjects were instructed what type of landing before jumping and the landings were monitored to assure that the entire foot contacted the force plate and that there was full heel contact. Subjects were allowed to rest at their convenience although most subjects had no difficulty completing the protocol without a break. All subjects performed the same nine jumps at each of the four tension levels in a predetermined, randomized order (Appendix E). EMG, motion
analysis, accelerometry, strain (in only four subjects) and force were collected for two
seconds - enough to capture both the counter movement and the landing. 1G trials were
conducted just as the ZGS trials, the only difference being an AMTI force plate was used
to collect the ground reaction force.

3.4 Instrumentation

Data collected during the zero gravity simulation and 1G trials required the
coordination of several systems and their corresponding transducers. Hardware was
manufactured and a LabVIEW (National Instruments, Austin, TX) program was written
to synchronize data from the motion analysis system, electromyography system, the tibial
and calcaneal accelerometers, a calcaneal strain gauge and the force plate. Data were
collected with a National Instruments analog to digital (A/D) board (Model AT-M10-16E-
10) in a Pentium personal computer. All instrumentation was coupled to the computer
through a junction box, with each transducer occupying a separate channel. Hardware
filters were wired into the system for the strain gauge and the acceleration data. A wide
band low pass filter, -3dB at 3.1 kHz and flat to 1.8 kHz was used for the strain
measurements. Both the acceleration and strain channels were subjected to an anti­
aliasing filter of 6dB per octave. Additionally, the gain was set to 10 for the acceleration
channels (Figure 3.9). Hardware supplied with the Bertec force plate used a two pole
filter with a cut-off frequency of 500Hz and the adjustable gain was set at 2 for these
experiments. The strain measurements channeled through the Capacitec amplifier included
a low pass filter with a 3.5 kHz cut-off frequency. Data were saved to an ASCII file with
each column representing one of six channels of data: strain, calf EMG, quadriceps EMG,
calcaneal acceleration, tibial acceleration and force respectively. All data were collected at
2400 Hz for two seconds with the exception of the motion analysis data which were
limited to 60 Hz.

In order to synchronize the motion analysis data which were collected on a
separate computer, a TTL pulse was sent from the A/D board in the PC to the Motion
Analysis VP320 video processor (Motion Analysis Corporation, Santa Rosa, CA) at the
Figure 3.9: Hardware Wiring Schematic
start of data collection. Similarly, a TTL pulse was sent to the PC from the video processor to indicate the start of the motion analysis data collection. The pulse was saved in the seventh column of the data file to allow for synchronization during data processing. During 1G trials, this pulse also triggered the AMTI force plate mounted in the floor.

3.4.1 Ground Reaction Forces

The Bertec force plate was used to record the ground reaction forces during the jumping trial. The force plate has the capabilities of recording six channels - three orthogonal forces and the corresponding moments. The output of a specific channel of the force transducer can be sensitive to another component of the load. When cross-talk is present, the readings from all six channels must be multiplied by a calibration matrix. One of the unique properties of the Bertec force plate is the ability to behave almost linearly (0.2% of the full scale output). This allows for a single channel of data to be collected without significant error due to cross-talk.

The Bertec force plate is a strain gauge based force transducer using the Wheatstone bridge configuration in the four corners of the plate. This force plate was used for all zero gravity simulation trials. Prior to data collection, the force plate was "auto-zeroed" to allow the electronic circuitry to balance itself. Only the vertical component of the force was collected from the Bertec force plate during jumping in the simulator. An AMTI (AMTI, Cambridge, MA) force plate was utilized during the 1G trials. Similar to the Bertec force plate, it is a strain gauge based force transducer. All six channels were collected on the AMTI plate and multiplied by the factory-supplied calibration matrix zero values which were manually chosen from the calibrated data files then subtracted from the force values to cancel any offset.

3.4.2 Acceleration

Lanyon (1987) has suggested that shock transients are a factor in the regulation of bone structure. To measure this phenomenon, Endevco accelerometers (Model 2250AM1-10, Endevco Corporation, San Juan Capistrano, CA) were attached on both the tibia and on the lateral calcaneus. These light weight accelerometers (mass 0.43 grams) were capable of recording up to 500G. The accelerometers were coupled to the skin over
the calcaneus and tibia via a cube of balsa wood (approximately the same size as the accelerometer) and oriented with the principle direction acting along the long axis of the body. Sensitivity in this direction was 9.75 mV/g and 9.79 mV/g for the calcaneus and tibia, respectively. Adhesive tape and pre-wrap secured the wood and accelerometer in place.

The potential for errors in the acceleration data may arise from the fact that accelerometers were mounted on the skin. Motion between the skin and the wood can cause artifacts to be introduced into the signal. Ideally, the accelerometer should be attached directly to the bone in the area of interest. This, however is not a practical solution. Ziegert and Lewis (1979) and Light et al. (1980) examined response of both bone and skin mounted accelerometers. No differences were found with low mass transducers (less than 1.5 grams). Therefore, using the current technology, no significant errors were expected in the acceleration data. Acceleration signals were used to assess the transmission of the force transients through the lower extremities and to ascertain the degree of shock absorption in the heel and ankle.

3.4.3 Strain Measurements

A new technology was utilized to measure the calcaneal strain. A Capacitec button transducer (HPB-75/156B-A-13-15-B-D, Capacitec Inc., Ayer, MA) is a capacitive based non-contact displacement transducer. The sensor amplifier design uses a linear capacitive reactance technique for converting voltage output to displacement. The disk shaped sensor, 1.905 mm in diameter, is surrounded by a ring guard approximately two times its size. Both are isolated from each other with a shielded conductor wire connected directly to the sensor (Figure 3.10). The ring guard surrounding the sensor minimizes distortion on the electrostatic field created between the sensor and the target and allows the sensor to maintain its linear range. The sensor measures displacement in the following manner. When positioned parallel to a grounded target, a capacitance is created which is proportional to the air gap between the sensor and the target. Most capacitive inductance techniques yield decreased sensitivity with decreasing sensor area - a disadvantage when small areas are required. When the area between these two capacitive surfaces is
decreased, the capacitance sensitivity is decreased. In contrast, the Capacitec transducer displays a large linear range with decreasing sensor size making it ideal for the current application.

![Capacitec Transducer](image)

**Figure 3.10:** Capacitec Transducer

To measure the calcaneal bone strain, the sensor is affixed to an aluminum post and mounted on an intracortical, stainless steel K-wire (1.98 mm diameter) in the calcaneus. Likewise, the target, formed by another grounded aluminum post, is attached to a second intracortical pin and a ground wire was connected to the extensometer set-up to protect the subjects during data collection (Figure 3.11). The linear range of the sensor is 1.270 mm. The sensor was calibrated to a 0.04 inch range. With the amplifier output of 0-10V, calibration coefficient of 0.004 inches/volt and a 0.001 volt resolution, the sensor can resolve strain up to $\pm 2 \times 10^{-6}$ inches. The gauge length for the Capacitec extensometer was 1.27 cm - based on the aluminum post configuration and the distance at which the pins were inserted into the heel, which yields a resolution of $\pm 3.81$ microstrain. The air gap is initially set at approximately 0.508 to 0.762 mm by adjusting the target which slides along a center post and is held in place by a set screw. Voltage levels from the Capacitec transducer were recorded during jumping and converted to bone deformation by the
calibration constant supplied by the manufacturer. In order to convert this value to strain, a "zero" reading was recorded from the sensor while the subject was non-weight bearing.

![Diagram](image)

**Figure 3.11: Extensometer Set-Up**

Since this technique has never been attempted during *in vivo* bone strain studies, it was anticipated that problems might arise due to the experimental set up. One of the main obstacles was the fact that the K-wires, the aluminum posts and the sensor itself have a certain mass. This may cause vibration in the pins when the subject lands and thus introduce artifact in the strain measurements. Using Castigliano’s Theorem, the deflection of the pins due to the acceleration acting on the system was determined (Appendix F). Additionally, if the pins loosen during the high impact jumping exercises, higher strains will be recorded than actually experienced by the bone. All means were taken to eliminate these problems by keeping the pins as short as possible and constantly monitoring the intracortical pins by checking for looseness and assessing the digital signal after each jump.

### 3.4.4 Kinematic Measurements

Kinematic measurements were included in the analysis to ascertain several factors. First, it was necessary to determine if landing kinematics were similar in the zero gravity simulator and on the floor. The variables which would indicate these differences include
the dorsiflexion/plantarflexion angle of the ankle and the flexion angle of the knee. Since pilot studies indicated that the knee flexion angle at landing varied between subjects, it is necessary to determine what effect knee flexion angle has on the force and other measured variables such as strain. Additionally, the kinematic measurements can give an indication of the impact velocity by calculating the second derivative of the position data of the heel marker. Jump height is determined by the maximum displacement of the hip marker. In the zero gravity simulator, the displacement of the springs in the gravity replacement system gave an indication of the variation of the force exerted on the subject during jumping and provided a means to validate the measured jump height.

Five spherical retroreflective markers were placed on the right leg of the subject at the second metatarsal head, the lateral malleolus, the posterior heel, the femoral epicondyle and the greater trochanter. Two markers were also placed on the inferior end of each of the GRS springs. The position of the markers was recorded by four cameras which were strategically located around the subjects in both the zero gravity simulator and on the floor. Both areas were calibrated and the position of all markers was recorded during jumping. Data were collected at 60 Hz (the maximum allowable for the system), transferred to the Sun SPARC station (Sun Microsystems Inc., Mountain View, CA) and synchronized with the other variables collected with the PC. Position data were tracked using the AMASS system to determine the x, y and z coordinates for all markers.

3.5 Data Processing

Having collected data from several different sources, numerous processing techniques were necessary to obtain the pertinent results for the experiments. LabVIEW programming was the primary means used to manipulate the data from the Bertec force plate, the accelerometers, EMG and the strain signal. Turbo Pascal (Borland International, Inc., Scotts Valley, CA), GX (in-house custom software) as well as LabVIEW were used on the motion analysis data. A brief discussion of the analysis procedures and the variables that they yielded follows.
3.5.1 Filtering and Splining Techniques

The first objective in data processing was to determine a filtering cut-off frequency and the type of filter that would be used on the data. Using the force profile from a jump on the floor, an FFT (Fast Fourier Transform) was applied to the signal. Few frequency components were found above 150 Hz; therefore, this value was chosen as the cut-off frequency for the force, acceleration and strain data. A fourth-order Butterworth filter with infinite impulse response cascade was utilized and had minimal effect on the overall magnitude of the signal - less than 1.7% of the peak signal was lost after filtering (Figure 3.12).

Splining was necessary for data which were not sampled at 2400 Hz. This included: the 60 Hz motion analysis data (including the displacement of the springs, knee and ankle joint angles, hip height and foot positions) and force plate data collected on the AMTI force platform at 1000 Hz in the 1G trials. The second derivative of the time series data were computed and used in a cubic spline interpolation routine found in the LabVIEW software. Although splining has the potential to introduce minor discrepancies in the data, it was necessary in order synchronize data captured and match time data for accurate analysis.

3.5.2 Ground Reaction Data

Ground reaction forces were captured from instrumentation in the zero gravity simulator and on the floor. For the Bertec force plate, no processing was necessary prior to utilizing it in the LabVIEW program. The AMTI force data, however, were submitted to several GX programs. The first program, ana2ts, converts the analog force file to time series data. The second program, cal_force, ran the data through a calibration matrix and allowed the zero value to be selected manually by positioning the cursor over a point in the flight phase (the values were calculated as the mean of the ten points before and after the cursor position). Finally, the program saved the z-component of the force in a file to be used later in the LabVIEW analysis program where it was splined to a 2400 Hz signal.
For all force data, the peak ground reaction force was determined during landing. Ground reaction forces were also used to compute the flight time (amount of time the subject is off the supporting surface) and the loading rate. Flight time was determined by the time between the start and end of the flight phase. The beginning of flight (or push-off) was defined as the point in time where the force dropped below 10 N after the counter-movement. Similarly, the end of the flight phase occurred when the force began to rise above the 10 N limit immediately following push-off. 10 N was chosen as the "threshold" level.

Loading rates for the force data were calculated over a period of ten time frames. However, due to the variability of the data, it was necessary to manually choose the loading rate window in the LabVIEW analysis program. Considering only the interval...
between the onset of landing (end of flight phase) and the peak force, the steepest portion of the curve was identified. A regression was performed over the ten points in the center of the steepest part of the curve to determine the loading rate.

3.5.3 Kinematic Data

Using the AMASS software (ADTECH, Aldelphi, MD), position data (x, y, and z coordinates) for each motion analysis marker were generated. The data were then smoothed, and filtered in GX using a custom designed program. The method proposed by Winter and Wells (1980) was used to determine the cut-off frequency for the kinematic data. Data for each marker were filtered over a range of cut-off frequencies and the goodness of fit between the raw data and the filter version was computed by the RMS deviation. The relationship between the residual and the cut-off frequency decreased asymptotically (Figure 3.13). The cut-off frequency which applied to the data files was determined by the value above which there was no significant difference in the RMS. For the kinematic data, all markers were observed and the value which was best suited all data was 9 Hz. A fourth order Butterworth recursive filter with a 9 Hz cut-off frequency and no phase lag was then applied to the position data in GX. Spikes in the data caused by incorrect classification of the markers were removed manually before filtering to assure a smooth curve.

Position data were used to determine the knee flexion angle, ankle dorsi/plantar-flexion angle, the jump height, impact velocity and spring displacement. In a Turbo Pascal program, the joint angles were calculated using the Law of Cosines and the Law of Sines. Adjustments were made for the offset of the ankle angle due to the placement of the markers on the foot and knee. Knee angle offsets were determined by averaging the angle during mid-flight phase of the jump and subtracting the value from knee angle data.

Jump height was calculated as the maximum displacement of the hip marker in the vertical direction (or horizontal in the zero gravity simulator). However, it was necessary to know the position from which the hip marker originated in order to get an accurate account of the overall distance the subject traveled. Therefore, the position of the hip marker recorder during the standing trial was used as the relative reference. A similar
method was used to determine the displacement of the springs during jumping in the zero
gravity simulator. Since each spring only had one marker attached to the inferior end,
displacement could only be measured relative to a pre-defined "origin". The origin chosen
in this case was the position of the spring marker at the instant of "take-off" (or the
beginning of the flight phase). Displacement was calculated as the length of the line drawn
from the origin to the position of the marker at each time frame.

![Graphs showing RMS Residual Between Raw and Filtered Data for Hip, Ankle, Knee, Heel, and Toe]

**Figure 3.13:** Relationship Between Filter Cut-Off Frequency and Residual. Performed on Kinematic Data
Once the initial contact of landing was determined from the force plate data, the impact velocity was calculated as the derivative of the position data for the heel marker just prior to landing (100 frames or 41.7 milliseconds before). The heel marker was chosen for this calculation because it was the closest to the supporting surface. This fact will introduce the least amount of errors; however, some variability may occur due to the landing styles - toe-heel versus flat-foot where the heel is contacting the plate almost precisely at the moment of landing. With a toe-landing there is a time lag between these two events.

3.5.4 Acceleration Data

Acceleration data recorded from the tibia and the calcaneus were adjusted for an offset by subtracting their respective readings during the standing trials in both 1G and simulated microgravity. Tibial and calcaneal shock were characterized by the peak value, time to peak and the power spectra. The shock transmission characteristics of the lower extremity were assessed in both the frequency and the time domain. The gain/attenuation profile of the shock as it travels up the leg was calculated from 0 to 150 Hz using Equation 3.8 (Lafortune and Lake, 1992):

\[ G(\omega) = 10 \log_{10} \left( \frac{\text{Power Spectra of the Tibial Acceleration}}{\text{Power Spectra of the Calcaneal Acceleration}} \right) \]  (3.8)

This profile was computed from the first 512 points after landing, yielding a frequency resolution of 4.88 Hz. The gain/attenuation profiles were also calculated for the relationship between the tibia - force and calcaneus - force. Additionally, in the time domain, the impact shock peak ratios were computed (tibia/calcaneus; tibia/force; and calcaneus/force).

3.5.5 Strain Data

Calcaneal strains were assessed by determining the peak compressive strain, peak tensile strain and the strain rate. Because of the high variability in the characteristics of the strain profile, it was necessary to manually determine the range of values over which the strain variables would be measured. The range which was chosen was approximately
between the time of landing and the time of peak or shortly thereafter. Once this range was chosen, the maximum and minimum values of strain were determined. The strain rate was calculated by performing a regression over the 10 points past the first peak (either minimum or maximum). This enabled the strain rate to represent the slope of the line between the compressive and tensile strain in the chosen range.

A zero value for the strain was necessary in order to determine the overall deformation in the bone. Zero values were recorded during the “zero” data collection when subjects were completely off the force plate and non-weight bearing on their instrumented leg. This value was subtracted off the raw strain (in volts). The data were then calibrated (multiplied by the calibration constant of 0.004 inches/volt) and divided by the distance between the two intracortical pins in order to express the values as strain. In the case when the zero reading on the strain gauge was not recorded properly (as in the case with Subjects 2 and 4), the value of strain in mid-flight was averaged over ten frames and used as the zero. The differences between these two “zeroing” methods were assessed by determining the variation in the output maximum and minimum strain.

In half of the subjects participating in the in vivo strain measurement studies, the zero value of the extensometer was accurately measured. In these two subjects, the differences between the strain when the zero value was directly measured and when it was calculated, using the average value of strain in mid-flight, was computed. The average offset value for both the maximum (tensile) and minimum (compressive) strain was $0.01433 \pm 0.02091$ strain. Since the calculated zero strain values were determined in the same manner for all subjects, the measured zero strains were adjusted using the offset value and calculated zero strain as applied to the two subjects which did not have the zero values properly recorded.

Deflection of both calcaneal pins according to Castigliano’s Theorem (Appendix F) was calculated and summed to yield the overall displacement measured by the strain transducer due to factors other than bone strain. The displacement was then used to adjust the strain values, decreasing the magnitude to solely reflect the compression in the bone produced by the impact forces during jumping exercises. However, this process only provided a rough estimate of the errors due to pin vibration. The deflection calculated by
Castigliano's Theorem is in the direction of the applied load. In the case of the calcaneal pin, however, the orientation of the pins is not necessarily in the direction of the force. Although the pins were positioned along the calcaneal trabeculae which are assumed to be oriented in the direction of force transmission through the bone, the force used in Equation 3.9, which is a product of the acceleration measured vertically, may not correspond to the direction of strain measurement. Therefore, further error is inherent in the strain which cannot be accounted for.

3.6 Cadaver Tests

Prior to conducting the zero gravity trials, a series of cadaveric drop tests were performed to assess the effect of high impact forces on the lower limb (Courtney et al., 1997). Cadaveric feet, dissected at the distal third of the tibia, were instrumented with accelerometers and strain gauges on both the tibia and the calcaneus in the manner same manner as the in vivo studies (Sections 3.3.2 and 3.3.4). The equipment used in the cadaver test was identical to that used in the in vivo trials. The proximal tibia was exposed and potted in an aluminum tube with bone cement. The limbs were secured in a custom-made drop test apparatus. This mechanism was equipped with crossbar which can be positioned at varied heights with additional weights to achieve a spectrum of peak loads and loading rates as the foot is dropped. When the crossbar was released, the foot impacts the force plate directly underneath. Position data were recorded using a linear variable displacement transducer (LVDT). Positions, tibial and calcaneal accelerations, strains and the vertical ground reaction force were recorded at 5000 Hz for 0.1 seconds subsequent to the foot contacting the force plate using a person computer, A/D board and a LabVIEW program (Section 3.4). Data were collected on ten limbs at five different heights with three weight levels added to the crossbar at each height. Peak load, acceleration and compressive strain at the tibia and calcaneus were determined for each trial. Data were filtered using a fourth order Butterworth filter with an infinite impulse response and a cut-off frequency of 150 Hz. Loading rates and strain rates were calculated by performing a regression over the ten points prior to the peak values. Impact velocity was determined by the derivative of the position data just prior to contact.
The cadaver data (peak forces, loading rates, peak accelerations, peak compressive strains and impact velocities) were compared to the results in the zero gravity simulated trials and the 1G trials. Additionally, the relationship between the external ground reaction forces and the calcaneal strains were determined in the cadavers and compared to the in vivo data. Due to the lack of the muscle actions at the ankle joint and on the bone, it was expected that both the peak compressive strains and the dependence of the strain on the impact force would differ from the in vivo trials. However, if the correlation between the force and strain values are similar, the relationship in the cadavers can be used to predict the bone strain based on ground reaction force alone. Therefore, tibial strains during jumping in the zero gravity simulator and on the floor can be estimated by this method since no strain gauge was implanted in the human subjects. For future studies, cadaver results can be used to derive bone strains for any activity without invasive means of implanting the extensometer.

3.7 Rheological Modeling

Rheological modeling provides a means to mathematically describe the stress/strain behavior of a material using elements that individually exhibit simple responses to applied forces and/or rates of loading. By combining these elements to form a network, complex behaviors such as viscoelasticity can be described (Black, 1988; Evans, 1973; Sedlin, 1965). A model of the heel, both fat pad and calcaneal bone, was developed using the ADAMS* software (Automatic Dynamic Analysis of Mechanical Systems, Mechanical Dynamics, Inc., Ann Arbor, MI). Each model (bone and fat pad) was developed separately and its response verified by data in the literature. Subsequently, the models were combined and the response of the “heel” to high impact forces was determined.

3.7.1 The Heel Pad Model

During walking and other impact activities, the heel is most commonly the first structure to contact the ground (Cavanagh and Lafortune, 1980). Therefore, the shock absorption capabilities of the human locomotor system are strongly influenced by the properties of the fat pad in the heel of the foot. Numerous studies have attempted to
characterize the mechanics of the human heel pad (Bennett and Ker, 1990; Cavanagh et al., 1984; Aerts et al., 1995). In vivo investigations have utilized pendulum or other impact experiments while in vitro tests were performed with isolated heel pads in mechanical testing machines. Differences in experimental set up and testing parameters make comparisons of the two types of tests difficult. Additionally, large discrepancies are evident in reported values of energy dissipation and stiffness. A mathematical model provides a more controlled method of studying the mechanical properties of the heel pad and to observe its effects on the rest of the locomotor system. A rheological model which would predict the non-linear, viscoelastic behavior of the human heel pad was developed in order to incorporate its properties into a hind foot model for high impact forces.

Figure 3.14 shows a schematic of the heel pad section of the model. Spherical balls with negligible mass were positioned 0.5 cm apart horizontally. Connecting arms were composed of elastic springs and viscous dampers. Arms 1, 4, 7, 8 and arms 2, 3, 6 and 9 were given stiffness values of $1.5 \times 10^4$ and $1.5 \times 10^5$ N/cm respectively and had unstrained lengths of 1.0 cm. The remaining arms consisted of Kelvin bodies - a spring and damper in parallel - with lengths of 0.5 cm in the unloaded state. The properties of these links were: arms 5, 10 and 13 - stiffness, $1.5 \times 10^4$ N/cm, damping, 3.0 N-s/cm; arms 11, 12, 14 and 15 - stiffness, $1.5 \times 10^4$ N/cm, damping, 435 N-s/cm. A compressive, sinusoidal force equal to $(600 \sin(2\pi t - \pi/2) + 1.0)$ was applied to the center nodes in the model.

An important point regarding the rheological model of the heel pad is that although the configuration of the springs and dampers are set up to yield responses similar to that of the heel pad tissue, these components do not have any physiological counterparts. One of the interesting features of the model is evident at Node 4 (between arms 11 and 12). As the node is displaced by the force, the action of Arms 11 and 12, both of which are Kelvin units, enables the model to behave in a viscoelastic nature becoming more stiff with increased force and exhibiting a hysteresis pattern on the return part of the cycle.
Simulation of impact on the heel pad with a sinusoidal force produced force-deformation curves comparable to those found in previous studies (Figure 3.15; Bennett and Ker, 1990; Aerts et al., 1995). A maximum deformation of 4.2 mm in the model fell well within the range reported by these investigators (1.9 to 6.3 mm). Comparisons for energy dissipation, however, are more difficult as experimental values show a wide discrepancy. Reported values were 95% during in vivo and in vitro impact tests and 30% during in vitro Instron tests. This large variation has been attributed to compliance in the proximal joints and soft tissues. Although the true energy absorption characteristics of the heel pad are unknown, adjustment of the parameters of the mathematical model will allow any energy absorption profile to be duplicated.
By eliminating the sources of variability between and within experiments, the rheological model of the heel pad is capable of analyzing the individual effects of stiffness, damping and loading rate. This ability becomes crucial when attempting to assess the role of the heel pad in the reduction of impact forces delivered to the body. More importantly, the rheological model of the heel pad must be able to exert its shock absorbing capabilities when acting in conjunction with the heel bone.
3.7.2 Trabecular Calcaneal Bone Model

The calcaneus was modeled as a cube of trabecular bone with a cross-sectional area of 20 cm$^2$ and the length of the sides equivalent to 5 cm (Figure 3.16).

\[
\text{Stiffness, } k = \frac{E \cdot A}{L}
\]

\[E = \text{Young's Modulus} \]
\[A = 20 \ \text{cm}^2 \]
\[L = 5 \ \text{cm} \]

**Figure 3.16:** Representation of Calcaneal Bone

Rheologically, bone can be represented by a combination of a Hookean spring and a Kelvin unit - a spring and damper in parallel (Figure 3.17; Sedlin, 1965). Similar to the heel pad model, a spring-damper model was developed using the ADAMs software. The model of the calcaneal bone consisted of a 2.5 cm Hookean spring with a spring constant $H$ in series with an equally long Kelvin unit of the same length having a spring constant, $K$ and damping coefficient, $\eta$. The magnitude of these parameters ($H$, $K$ and $\eta$) dictates the overall stiffness of the heel bone and therefore, it was important to determine the magnitude which would best describe the properties of the heel.

Sedlin (1965) stated that for a constant rate of deformation, the stiffness of the bone is related to the constant in the rheological model (Equation 3.9).

\[
\frac{HK}{H + K}
\]

**Equation 3.9**

At rapid rates of deformation the effect of the damper will cause the Kelvin unit to become infinitely stiff and not deform. Therefore, the stiffness of the system in this case is solely defined by the constant, $H$. 

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Since trabecular bone stiffness is quite variable and it was not possible to obtain accurate densitometric measure from the subjects, it is necessary to use data in the literature to gain an estimate for the stiffness of the calcaneal bone. Carter and Spengler (1982) related the modulus of elasticity of bone \( E_c \) to its density and strain rate:

\[
E_c = 3790 \times \text{(density)}^3 \times \text{(strain rate)}^{0.06}
\]  

(3.10)

A value 0.6 g/cm\(^3\) was chosen as the apparent density of the calcaneal bone. While 0.6 g/cm\(^3\) is on the high end of the values reported for the density of trabecular bone, it was felt that a higher value would help account for the thin shell of cortical bone on the outer surface of the calcaneus. Selecting a high strain rate of 0.1 strain/s, the value of \( E_c \) was computed. At the given strain rate and density, the value for the stiffness according to Equation 3.10 was 165,047 N/cm. This value was assigned to the constant, \( H \) and used in Equation 3.9 to determine \( K \) equal to 518,600 N/cm. The value of the damper was initially set to 2.44 MPa/s as described by Smith and Keiper (1965). After several simulations two factors were obvious. The damper had little or no effect on the bone strain and the overall bone strain was much too high in terms of the range seen in the physiological measurements. Because of the inverse relationship found between the strain and the damping coefficient (as damping increases, strain decreases), it was decided that \( \eta \) would be doubled to 5.0 MPa/s.
A mathematical derivation of the bone model was performed for two purposes (Appendix G). First, it was necessary to assess the ability of the rheological model to predict realistic bone strains with a given force input. Additionally, it was also used as a mean to verify the ADAMS model of the bone. Peak magnitudes of the force were set to 3,000 N in a 1Hz sinusoidal signal and the strain was calculated using the Equations outlined in Appendix G. The stiffness of the bone was determined by the mathematical force-deformation characteristics and coupled with the geometric characteristics of the calcaneal trabecular bone, the modulus of elasticity was calculated from the following relationship:

\[ E_m = kL/A \]  

(3.11)

where \( k \) is the slope of the load-displacement curve, and \( A \) and \( L \) are the area and length of the model of the calcaneal bone, respectively (Figure 3.16). For the bone, the rheological model predicted stiffnesses, \( E_m \), compared favorably with those found using the Carter and Spengler relationship (Equation 3.10) at a given value of \( \eta \). However, an inverse relationship was found between the two estimated moduli as the damping coefficient increased.

In ADAMS, a compressive sinusoidal force, \( F \), was applied to the model in order to obtain load-deformation profiles, and rates of compression.

\[ F = M*(\sin(100 \, t - 12) + 1.0) \]  

(3.12)

The peak strain was determined and the magnitude was identical to that calculated by the mathematical model (0.28% strain).

3.7.3 Combining the Heel Pad and the Bone Model

The ultimate goal in designing the rheological model of the foot was to simulate the response of the lower extremity to high impact loading similar to that seen in the experimental jumping exercises performed in 1G and in simulated microgravity. With both the bone model and the heel pad yielding responses similar to measured values of the
individual material, the two were combined and grounded at the top of the bone (Figure 3.18). The bone was attached at Node 4 of the heel pad in order for the heel pad to have the greatest viscoelastic effect on the force acting on the bone.

Accelerating Block

Figure 3.18: Rheological Model of the Heel - Calcaneal Bone and Heel Pad Combined

Since the desire was to study the effect of impact forces on the bone, it was decided that a sinusoidal force would not adequately simulate jumping exercises. To create a more impulsive force, an accelerating block was connected to the heel pad
through a series of springs (Figure 3.18). Several modifications were necessary to the heel pad model for stability when attached to the bone. In the heel pad alone (Figure 3.14), symmetry was present with equal force being applied to the corresponding nodes on either side (Nodes 4 and 9). In the hind foot model, the force applied at the bottom, however, is slightly different than the one experience by bone. This asymmetry cause the model to be very sensitive and become unstable relatively quickly. Therefore, two new spring were added to the model with a very high stiffness of $1.5 \times 10^7$ (shown in the heavy lines in Figure 3.18). These springs provided stability to the lower half of the heel pad as it was displaced vertically. It also eliminates some of the viscoelastic effect at Node 9 (central node - similar to Node 4). The stiffness of the outer arms was increased to $7.5 \times 10^4$ N/cm for the same reason.

During simulations in ADAMS, the accelerating block was given a mass and an initial velocity which would be capable of delivering a high impact force to the hind foot. The force of the block created a reaction force in the connect springs which in turn deform the fat and bone respectively. Simulations were run with four different magnitudes of block mass (2.5, 5.0, 7.5 and 10.0 kg) at four initial velocities (25, 50, 75 and 100 cm/s). In order to adequately compare the model with the experimental data, values of initial velocity and mass of the block were chosen to produce force magnitudes similar to the ground reaction force measured in the zero gravity simulator and in 1G.

For all combinations of mass and initial velocity, the force in the block, the force in the Hookean spring of the bone, the deformation of the fat pad and the strain in the bone were measured in the first half of the cycle as the force approaches its peak. The deformation in the fat was calculated as the displacement between Node 4 and Node 9 divided by the initial distance of 0.968 cm. This is only half the distance of between Node 4 and 9. Essentially, only the top portion of the heel pad is having a viscoelastic effect on the system. The bottom portion was stiffened by the addition of the middle springs and only supplies stability to the model. Since the heel pad model is symmetric, it is feasible to assume that the top half has the same effect as the whole. The bone deformation was calculated in a similar manner by determining the displacement between Node 4 and the
grounded end of the bone (Node 16) and dividing by the initial distance of 5 cm. The stiffness in the bone was determined by the slope of the force deformation curve and strain rate was calculated as the slope of the strain versus time curve in the first part of the cycle.

3.8 Summary

Combining all of the variables discussed above, a complete understanding of the shock transmission through the lower extremities during jumping can be obtained. Furthermore, the rheological model has the ability to accurately predict the physiological occurrences in the bone in response to high impact force inputs. These data should help to determine the effectiveness of jumping in creating an environment of high strain and force which enables the osteoregulatory response to eliminate bone resorption during space travel.
CHAPTER 4

RESULTS

The intent of the present study was to gain insight into the biomechanical aspects of jumping in simulated microgravity and the possible effectiveness of this exercise as a countermeasure to prevent bone demineralization which occurs during most space flights. The aspect of the jump which was primarily examined was the landing, with three landing types utilized: two-footed toe-heel landings, one-footed toe-heel landings and two-footed flat-foot landings. The variety in landing types allowed for a spectrum of responses with respect to ground reaction forces, tibial and calcaneal accelerations, calcaneal bone strains, ankle and knee kinematics and electromyography to be measured during jumping exercises in both simulated microgravity and 1G. Comparisons were made for all conditions to ascertain the differences among the responses at the five spring tensions as well as between simulated microgravity and 1G. After a brief review of the performance of the zero gravity simulator, the results presented in this section will address the aims and hypotheses stated in Chapter 1.

4.1 Zero Gravity Simulator

The zero gravity simulator was designed to negate the effects of gravity on the lower limb segments while the subject was suspended in a supine position. To do so, it was necessary that the tension (and therefore the length) of the cord remain constant throughout the duration of the jump. However, during the countermovement of the jump, bent knees caused a decrease in the length of the cords by approximately 18.6%. On the landing, the deviations were smaller amounting to only a 11.8% decrease in length. Changes in lengths were measured with respect to the location of the center of gravity in
the standing posture. Since it was difficult to maintain a uniform horizontal position in the simulator, deviations in the center of gravity position were noted throughout the entire jump. While these values are relatively small and had no apparent effect on the jumping mechanics, the fact that the cord tension is not constant is a limitation in the design of the zero gravity simulator. In order to circumvent this design flaw, the cords could be mounted on the ceiling through a series of pulleys (Davis, 1991) which would maintain the necessary length through the entire jumping phase.

Contributing to the decrease in the cord tension is the "pendulum effect." The pendulum effect occurs as the subject jumps out away from the force plate. The cords, which are aligned with the center of gravity of the thigh and calf during standing are displaced from the vertical alignment proportional to the distance the subject moves away from the supporting surface (the jump height). The force acting in the vertical direction then becomes a component of the force in the cords. The average jump height in the zero gravity simulator was 25.44 cm. This creates a 4.77 degree angle between the cords and the vertical. Therefore, the vertical component during flight is decreased by less than 1%. From these data, it can be concluded that the "pendulum effect" has a negligible contribution on the overall vertical force in the cords. Additionally, this information shows that a ten foot high ceiling is an adequate distance for a supine suspension system.

The gravity replacement system (GRS) of the zero gravity simulator was calibrated to tether the subject to the supporting surface with a force equivalent to a percentage of the subject's body weight. The length of each spring necessary to attain this force level was determined by the force-displacement characteristics of the spring as well as the angle at which the GRS was attached to the body and the subject's body weight. The ability of the gravity replacement system to achieve the desired force was measured by ground reaction force during quiet standing. The equilibrium force measured during standing was expressed as a percentage of body weight and compared to the anticipated value for each of the four tension levels (45, 60, 75 and 100% of body weight) as dictated by the spring length. As demonstrated by Table 4.1, the difference between the expected and measured value increases as the tension level increases. This relationship can be attributed to three
factors. First, the higher tension levels required the springs to be stretched to almost twice their resting length and therefore, more difficult to achieve. Additionally, at the higher tensions, the springs most likely were not able to maintain the large extensions due to slip in the ropes and knots and in the springs themselves. Another factor which contributes to the error between the expected and measured values is the subject's posture during the equilibrium force measurements and during the lengthening of the springs. Although subjects were instructed to keep their knees straight during both of these conditions, any bend in the knees would cause the measured force to be decreased.

<table>
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<tr>
<th>Expected Tension Level (%)</th>
<th>Average Measured %BW</th>
<th>Average Difference (%)</th>
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<td>6.07</td>
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<td>60</td>
<td>49.98</td>
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Table 4.1: Percent Body Weight at Four Spring Tensions in the Zero Gravity Simulator

Another aspect that can be explored with respect to the performance of the zero gravity simulator is the overall displacement of the springs and the jump height. Both the jump height and the spring displacements were measured from the displacement data of the reflective markers. Since only one marker was placed on the springs due to experimental constraints, the displacement of the spring was measured with the position at the beginning of the flight being used as the origin. Although the springs of the gravity replacement system were oriented in a vertical direction and attached to the horizontal subject through a series of ropes and pulleys, it was expected that the movement of the springs would reflect the horizontal distance covered during jumping. As the subject left the supporting surface, the inferior portion of the springs, upon which a reflective marker was mounted, were pulled towards the ground. The displacement of each spring was measured and compared. Since jump height was significantly different for each tension
level ($p < 0.05$), the comparison between spring displacement and height was considered separately for each tension level.

At all tension levels, Spring 1 was displaced approximately 10% further than Spring 2 (Table 4.2). The difference in the spring displacement and the jump had no apparent trend with the 75% tension level showing the greatest similarity followed by 100, 60 and 45%, respectively. The differences were also equally split with regard to the magnitude of the jump height versus the spring displacements. Errors in spring displacements could be attributed to the fact that the retro-reflective markers mounted on the inferior end of the springs were on the very edge of the field of view of the cameras. This may have introduced a non-linearity in the position data of the spring markers affecting the measured magnitude of the displacement. The fact that no trend exists highlights the variability in the jump height data. If you look at the displacement of several subject's data individually (Table 4.2), a definite trend of decreasing spring displacement with increasing tension level is evident. Averaging across subjects eradicates this trend because the jump heights were so different.

Because the springs of the gravity replacement system are stretched during jumping, the force pulling the subject toward the supporting surface increases with jump height. Additionally, as the tension level increase, the force rises proportionally. This creates a unique situation in the gravity simulator as opposed to jumping exercise on the ground. In 1G, the gravitational force which must be overcome to gain height is constant and has a uniform effect on the impact velocity of the foot at landing. In the simulator it increased the further the springs were stretched.
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<th>Spring 2 Displacement (cm)</th>
<th>Jump Height (cm)</th>
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Subject 5

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<td>26.10</td>
<td>36.18</td>
<td>23.33</td>
</tr>
<tr>
<td>75%</td>
<td>25.40</td>
<td>26.62</td>
<td>22.85</td>
</tr>
<tr>
<td>100%</td>
<td>21.66</td>
<td>18.45</td>
<td>18.81</td>
</tr>
</tbody>
</table>

Subject 6

<p>| | | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
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<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>45%</td>
<td>27.79</td>
<td>35.30</td>
<td>24.79</td>
</tr>
<tr>
<td>60%</td>
<td>28.29</td>
<td>28.42</td>
<td>25.16</td>
</tr>
<tr>
<td>75%</td>
<td>27.95</td>
<td>24.33</td>
<td>25.74</td>
</tr>
<tr>
<td>100%</td>
<td>28.08</td>
<td>17.76</td>
<td>26.10</td>
</tr>
</tbody>
</table>

Table 4.2: Difference Between Jump Height and Spring Displacement for Six Subjects
Average Spring Displacement and Jump Height ± Standard Deviation

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4.2 Biomechanical Aspects of Jumping

The biomechanics of jumping were assessed by measuring the ground reaction forces, joint kinematics and shock transmission through the lower limb via tibial and calcaneal accelerations. The results will give insight into the ability to jump in simulated microgravity and also highlight the differences between zero gravity and 1G exercises. The latter is a critical determinant for developing exercise protocols in space which will be effective in combating space flight induced osteoporosis.

4.2.1 Peak Ground Reaction Forces

The vertical component of the ground reaction force was measured in the zero gravity simulator and in 1G. As illustrated in Figure 4.1, the ground reaction force rises in the countermovement phase of the jump and returns to zero during the flight phase. During landing, the force rises rapidly to the peak as the body is decelerated followed by a more gradual decrease in force. It is important to note that while most vertical ground reaction curves exhibited the characteristics shown in Figure 4.1, there was variation both between and within subjects. Results reported in the literature, identify two prominent peaks in the force profiles upon landing (Mizrahi and Susak, 1982; Dufek and Bates, 1990; Schot, Bates and Dufek, 1994). While peaks are evident in the present data, particularly in the flat-footed landings (Figure 4.1c), they are not as prominent and only the overall maximum force was considered for analysis. A possible cause of the disparity in the shape of the curve can be attributed to the different sampling rates used in the respective studies.

Previous studies have reported the peak force during landing as a percentage of the subject's body weight. This normalization process allows for the data between subjects to be compared and eliminates the dependence on body weight. In the present study, when the force was normalized to the equilibrium, standing value, there was little correlation between the peak force in landing and body weight (0.20, 0.41 and 0.24 in the ZGS, 1G and combined conditions, respectively). If the force were to be normalized by dividing by body weight, the correlation between normalized force and body weight should decrease.
Figure 4.1: Representative Ground Reaction Force During Jumping
(a) Two-Foot Toe-Heel Landings, (b) Two-Foot Flat-Footed Landings, (c) One-Foot Toe-Heel Landings
indicating no interdependence between variables. However, when body weight is used for normalization, the correlation increased to 0.42 for both the ZGS and ZGS and 1G combined conditions. Therefore, no normalization techniques were used for these data.

The analysis of variance (ANOVA) performed on the force data found a significant effect ($p<0.05$) due to the landing type. When the three landing types are considered (Type I - two foot toe-heel, Type II - two-foot flat-footed, and Type III - one-foot toe-heel), Type II landings (2631±663 N) are significantly higher than either of the toe-heel landings (Type I 1902±607 N and Type III 2394±754 N, Figure 4.2).

![Figure 4.2: Peak Force By Landing Type. Type I - Two Foot Toe-Heel, Type II - Two-Foot Flat-Footed, and Type III - One-Foot Toe-Heel](image)

A possible cause of this phenomenon can be attributed largely to the knee motion at landing. Knee flexion angle reached an average value of 17.2 and 12.7 degrees for the Type I and Type III toe-heel landings and 19.2 degrees for Type II. Initially, it was assumed that the knee angle would be greater in the toe-heel landings. However, the statistical results of the magnitude of the knee angle at landing showed knee flexion was significantly higher at impact for flat-footed landings compared to one-footed landings and no statistical differences were found between the Type I and Type II landings. This suggests that the motion of the knee following landing plays a role in the shock absorption and the stiffness of the body as it is decelerating upon landing. Figure 4.3 clearly
Figure 4.3: Landing Force and Knee Flexion Angle
Heavy line represents vertical ground reaction force and thin line represents knee angle. (a) Type I Landing; (b) Type II Landing; (c) Type III Landing
illustrates the fact that the knee continues to bend even after initial contact. In the Type II landings, however, the flexion is abruptly stopped after about 0.1 seconds while it continues for almost a quarter of a second during the toe-heel landings. Besides the shortened duration of knee flexion, the flat-footed landings also have a small range over which the knee is flexing after impact. McMahon, Valiant and Frederick, (1987) demonstrated a decrease in the overall stiffness of the body with an increase in the knee flexion angle by modeling the body as mass-spring system. By decreasing the stiffness, the impact force is decreased as well as the transmission of the shock transient from the foot through the body. These results in Figure 4.3 clearly illustrate the decreasing stiffness by increasing knee flexion angle and emphasizes the role of the joint motion in reducing impact force.

In the zero gravity simulator, the peak force during landing was not dependent on the spring tension level. No significant differences were found for peak force in any of the four gravity levels (p>0.05). This was a surprising outcome considering that the equilibrium weight increased with the increasing spring tension. However, the height that the subject could attain, decreased. The jump height is highly negatively correlated with the peak force (-0.717) and was significantly lower at each increasing tension level (Figure 4.4). Several explanations can account for the difference in heights. As the tension level increases, the resistance in the spring rises. This causes a more rapid decrease in the vertical lift-off velocity of the body's center of mass at higher tensions. Muscle force also plays a large role in the ability of the subjects to resist the spring. For this reason, several subjects had difficulty gaining height at the 75 and 100% levels. A possible explanation can be found in the work of Loveland et al. (1992) who, using a optimal control model, found similar effects on jump height at increasing higher gravitational levels. The authors also noted that the decreases in jump height was due to the velocity of the center of gravity at lift-off. While this parameter was not measured for the present data, the fact that the force opposing the jump is increasing with the tension levels would seemingly decrease the initial velocity.
For all peak forces in the zero gravity simulator, a comparison between tension levels showed differences between 10 and 46 N. Paired t-tests showed that a difference of approximately 214 N could be detected with a power of 80% in the group of twelve subjects (Appendix B). Therefore, it can be concluded that the peak force levels in the zero gravity simulator were all within 214 N of one another. The fact that this difference is only approximately 10% of the peak values seems to indicate the undue necessity of various tension levels - especially those which caused discomfort to the subjects - to achieve higher forces.

Figure 4.4: Impact Forces and Flight Times at the Four Tension Levels in the Zero Gravity Simulator and in 1G

While there were no apparent differences in force within the zero gravity simulator, there was a significant difference between the ZGS and 1G. Peak impact forces were greater in 1G (Figure 4.4). Using the previous argument, one would expect that the jump height is also significantly higher in 1G. This fact, however, only holds true for the 75 and 100% tension levels. At 45 and 60% body weight, the jump height was not significantly higher in the ZGS than in 1G. It can therefore be assumed that the magnitude of the equilibrium standing force contributes to impact force at landing. In the zero gravity simulator, the equilibrium force was never as high at the subject's body weight in 1G. This is a limitation of the zero gravity simulator which in the current design cannot be
overcome. If the equilibrium force value did reach 100% body weight in the simulator, it is uncertain whether the peak impact force would achieve the levels found in 1G. The fact that the knee flexion angle was significantly higher at impact in the ZGS suggests that the impact force will remain lower.

The force profile was also used to determine the flight time (Figure 4.5). The beginning of the flight phase was defined as the point at which the force fell below 10 N. Similarly, at the end of the flight phase, the force rose above this threshold. The flight time was statistically less with increasing spring tension in the zero gravity simulator. Flight time in 1G was statistically different from all ZGS tension levels falling in the middle range.

![Figure 4.5: Flight Times at the Four Tension Levels in the Zero Gravity Simulator and in 1G](image)

### 4.2.2 Loading Rates

Loading rate is defined as the time rate of change of the force during the landing phase of each jump. Loading rates were calculated by visually examining the force profile after impact (Figure 4.6) and determining the interval over which the slope of the force-time curve appeared the greatest. The loading rate was calculated using a linear regression over the ten points in the central region of the selected interval (gray area of Figure 4.6). The upper end of the interval was always considered as the index of the peak force. The minimum index of the force interval was determined manually and therefore varied from subject to subject. This variation was necessary in order to supply the
LabVIEW program (which performed the regression) with the ten points at the steepest part of the curve and should have no effect on the variation of the calculated loading rate.

![Force Profile from Impact to Peak Load](image)

**Figure 4.6:** Force Profile from Impact to Peak Load
Gray Area Represents Portion of Curve Used to Calculate Loading Rate

Loading rates were significantly different for the three landing types with Type II greatest (Figure 4.7). No significance was found for loading rates at the different spring tensions.

![Loading Rates by Landing Type](image)

**Figure 4.7:** Loading Rates by Landing Type
The magnitude of the loading rate ranged between 221 and 427 kN/s (Figure 4.7) and was significantly different for landing type (Type II > Type III > Type I). The difference in the loading rates is evident when viewing the impact force profiles for the three landing types (Figure 4.3). The force during flat foot landings (4.3b) is relatively sharp and moves directly to the peak in a shorter period of time. The toe-heel landings are more gradual and do not achieve the same magnitude. The differences in loading rates between the two toe-heel landings are most likely attributed to the fact that the entire weight of the subject is being decelerated entirely on one leg at impact during the Type III landings. In the Type I landings, the weight is distributed to both feet, one of which is on the supporting platform where the force is not recorded. The ANOVA performed on loading rate showed no significant differences between spring tensions including the comparisons between the four ZGS levels and 1G. The differences between the loading rates at the varied spring tensions extended from 1.7 to 2.8 kN/s. Paired t-tests showed a mean difference of 67 N would be detected with a power of 80% with the experimental data (Appendix B). Therefore, it can be assumed that loading rates at all tension levels were within 67 kN of one another. It has been hypothesized that the loading rate during impact exercises is a critical component in skeletal homeostasis (Cavanagh et al., 1992). Therefore, in the current investigation, the fact that the loading rates are within 20% regardless of whether it is measured in simulated microgravity or 1G is an important finding. It suggests that the magnitude of loading can be achieved during jumping on Earth can be obtained in space. Furthermore, if loading rate is a substantial determinant for bone strain, jumping would provide means for the skeleton to achieve osteogenic stimuli in space it receives in 1G.

4.2.3 Kinematics and Motion Data

Several different variables were measured using the motion analysis data derived from the position of the reflective markers on the right leg. These included: knee and ankle angle, impact velocity, and jump height. Each segment of the right leg was defined by a vector between the two markers. The thigh vector, from the knee marker to the greater trochanter marker, and the shank vector, from the knee marker to the lateral
malleolus marker, were used to determine the knee angle (Figure 4.8). Likewise the, shank vector and the foot vector, from the heel marker to the metatarsal marker were used to calculate the ankle angle (Figure 4.8). The knee angle was positive in flexion and negative in extension where an angle equal to zero represented a straight leg. It is important to note that the measured angle is not the included angle but rather, its complement. The convention for the ankle angle reported a zero angle when the foot vector was perpendicular to the shank vector. In dorsiflexion, the ankle angle becomes increasingly positive and in plantarflexion, negative.

![Diagram of knee and ankle angles](image)

Figure 4.8: Convention for Defining Knee and Ankle Angle
Filled Circles Represent Markers At The Greater Trochanter, Lateral Femoral Epicondyle, Lateral Malleolus, Posterior Heel And Second Metatarsal Head

Representative ankle and knee motion during jumping in the zero gravity simulator and in 1G are found in Figures 4.9, 4.10, 4.11 and 4.12, respectively. The start of the flight phase is evident in all angle data. At the end of the countermovement phase knee
angle reaches a minimum value of approximately zero indicating a straight leg in flight. For the ankle angle, Type I and Type III landings follow the same general pattern - peak plantarflexion is reached during flight and followed by dorsiflexion upon impact.

Interestingly, the Type II landings show a slight different pattern with the ankle beginning to dorsiflex shortly after take-off. This seems to indicate the subject's preparation for the flat-foot landings. In Figure 4.9 and 4.10, this occurs between 0.3 and 0.5 seconds indicating the beginning of the flight phase. For the 1G jumps (Figures 4.11 and 4.12), the flight phase begins between 0.4 and 0.6 seconds following the onset of the jumping activity. The time are different for each angle profile due to the fact that they are derived from different jumps and therefore are not synchronized. Impact occurred at approximately 0.9 seconds in simulated microgravity and 0.8 seconds for the Type II and III landings and 1.0 second for the Type I landings in 1G.

![Figure 4.9: Ankle Angle During Jumping in the Zero Gravity Simulator](image)

Figure 4.9: Ankle Angle During Jumping in the Zero Gravity Simulator
Figure 4.10: Knee Angle During Jumping in the Zero Gravity Simulator

Figure 4.11: Ankle Angle During Jumping in 1G

Figure 4.12: Knee Angle During Jumping in 1G
The magnitude of the knee flexion angle was determined at impact during jumping exercises (Table 4.3). The amount of knee flexion during one-footed toe-heel landings was significantly lower than either Type I or Type II landings \( (p<0.0001) \). The stiffer knee in Type III landings can be attributed to the fact that the subjects' entire weight needed to be received on one leg and increased flexion at the knee would create an unstable base of support eventually leading to collapse if the angle were too high. Furthermore, most subjects were unstable during one-footed landings and had difficulty maintaining balance upon impact. Knee angles were also significantly decreased in 1G compared to the zero gravity simulator, not a surprising conclusion in light of the flexed posture assumed in the simulator. Using the paired t-tests (Appendix B), it was determined that all values of knee angle at impact for all condition levels were within four degrees of one another.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Knee Angle at Impact (deg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Landing</td>
<td></td>
</tr>
<tr>
<td>Type I</td>
<td>17.2</td>
</tr>
<tr>
<td>Type II</td>
<td>19.2</td>
</tr>
<tr>
<td>Type III</td>
<td>12.7**</td>
</tr>
<tr>
<td>Spring Tension</td>
<td></td>
</tr>
<tr>
<td>ZGS 45%</td>
<td>20.2</td>
</tr>
<tr>
<td>ZGS 60%</td>
<td>18.0</td>
</tr>
<tr>
<td>ZGS 75%</td>
<td>19.0</td>
</tr>
<tr>
<td>ZGS 100%</td>
<td>18.3</td>
</tr>
<tr>
<td>1G</td>
<td>6.3**</td>
</tr>
</tbody>
</table>

Table 4.3: Knee Flexion Angle at Impact
** Indicates Significant Difference

The peak angles measured during the landing phase of the jump are illustrated in Table 4.4. Following impact, the knee began flexion and continued until achieving a peak. No significant difference were found for peak knee angle in the zero gravity simulator; however, the knee angle in 1G was significantly higher. Additionally, the maximum flexion
angle during landing toe-heel on two feet was significantly higher. Knee angular velocities range from 85 degrees/second to 126 degrees/second and followed the same trends as the peak angle during landing (Table 4.5). Peak ankle angle during landing ranged between 22 and 31 degrees of dorsiflexion. For toe-heel landings, the range of dorsiflexion following impact was much greater compared to the Type II landings in which subjects began to prepare for the flat-footed landing by dorsiflexing early in flight; however, the mean peak value was not significantly different from Type II landings. This is evident when observing the decreased angular velocity of the ankle (Table 4.5) indicating the limited range of motion.

<table>
<thead>
<tr>
<th>Ankle Angle</th>
<th>Knee Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Average</strong></td>
<td><strong>Std Dev</strong></td>
</tr>
<tr>
<td>Landing 1</td>
<td>27.50*</td>
</tr>
<tr>
<td>Landing 2</td>
<td>22.88</td>
</tr>
<tr>
<td>Landing 3</td>
<td>24.16</td>
</tr>
<tr>
<td>Tension 1</td>
<td>25.20</td>
</tr>
<tr>
<td>Tension 2</td>
<td>23.85</td>
</tr>
<tr>
<td>Tension 3</td>
<td>23.30</td>
</tr>
<tr>
<td>Tension 4</td>
<td>22.09</td>
</tr>
<tr>
<td>1G</td>
<td>31.64**</td>
</tr>
</tbody>
</table>

Table 4.4: Mean Peak Ankle and Knee Angle During Landing
* Significantly greater than Landing 2;3; ** Significantly greater than Tensions 1-4
+ Significantly greater than Landings 2,3; ++ Significantly greater than Tensions 2,4

<table>
<thead>
<tr>
<th></th>
<th>Ankle Angular Velocity (deg/s)</th>
<th>Knee Angular Velocity (deg/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Landing 1</td>
<td>218.41</td>
<td>222.34</td>
</tr>
<tr>
<td>Landing 2</td>
<td>132.05</td>
<td>219.65</td>
</tr>
<tr>
<td>Landing 3</td>
<td>104.66</td>
<td>200.26</td>
</tr>
</tbody>
</table>

Table 4.5: Mean Knee and Ankle Angular Velocities for Three Landing Types
Impact velocity was determined by regressing over 100 points prior to impact of the heel position data. ANOVA results showed a significant interaction in this variable for spring tension and landing type. Therefore all combinations of these two variables were compared and significance level assessed using an alpha value equivalent to the test significance level, \( \alpha \), divided by the number of comparisons. When each landing type is considered individually, the effect of each tension level on the impact velocity can be observed. For Type I landings (Figure 4.13), the impact velocity was significantly greater in 1G than in the zero gravity simulator (\( p<0.001 \)). In the zero gravity simulator, impact velocity was greater for the lowest tension. This again, can be credited to the increased jump height at 45% of the subject's body weight. Similarly, for the flat-footed landings (Figure 4.14), the impact velocity was significantly higher in 1G (\( p<0.001 \)). Impact velocities at Tension 1 were significantly greater than Tensions 3 and 4 as well as the impact velocity at Tension 2 being greater than Tension 4. One footed landings produced significantly greater impact velocities at Tension 5 than any of the ZGS tensions (Figure 4.15).

![Impact Velocity for Type I Landings](image)

**Figure 4.13:** Impact Velocity for Type I Landings
Tension 1 - ZGS 45%; Tension 2 - ZGS 60%; Tension 3 - ZGS 75%;
Tension 4 - ZGS 100% (± Standard Error)
* Denotes Significance
4.2.4 Acceleration

Both tibial and calcaneal accelerations were recorded during jumping exercise in the current study. A tremendous amount of information regarding the shock absorbing capabilities of the lower extremities can be derived from this information, as well as the body’s shock transmission characteristics under the different impact conditions. Acceleration data were assessed in both the time and frequency domains and were classified by the peak values during impact in the time domain and the gain/attenuation
profile in the frequency domain. The gain/attenuation describes the spectral modifications of the shock wave as it travels through the body from the imparting surface to the calcaneus and to the tibia. Initially, the frequency analysis was limited to 10-150 Hz bandwidth with a resolution of 4.6875 Hz. An ANOVA analysis was performed on both acceleration signals to determine the effect of the tension level and the landing type on the peak signal. Acceleration data from Subject 1 was not included in the analysis due to difficulties in obtaining clean signals.

Initial attempts were made to filter the acceleration data at 150 Hz, the cut-off frequency used when filtering force and strain signals. However, it appeared that the peak magnitudes were severely truncated at this frequency, especially for the calcaneus. An FFT of the calcaneal acceleration signal was obtained and, as evident from Figure 4.16, there is significant content over 150 Hz - extending as far as 600 Hz in some cases. It was apparent that an increase in cut-off frequency was necessary in the Butterworth filtering processing for the acceleration. The problem with selecting a magnitude for the cut-off frequency was the large variability both between and within subjects in the signal’s frequency content that often extended up to the 400-600 Hz range. On further inspection, however, the differences in the peak values using 400 Hz and 600 Hz were not significant and 600 Hz was used as the cut-off frequency in the Butterworth filtering of all acceleration signals. Since the data were passed through an anti-aliasing filter at 500 Hz, the new cut-off frequency appeared reasonable.

![Figure 4.16: FFT of Calcaneal Acceleration Signal](image-url)

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Representative tibial acceleration during jumping is illustrated in Figure 4.17. Tibial acceleration rises rapidly at impact and, on most traces, oscillations were present after the initial peak. Peak magnitudes of this parameter varied from 30.2 to 49.2 g (Table 4.6). ANOVA results showed significant differences for both spring tension and landing type. Most notable is the fact that tibial acceleration is significantly higher in 1G than in simulated microgravity. Additionally, flat-footed landing produced higher acceleration than either the two-footed toe-heel landings or one-footed toe-heel landings. The reason for these differences has already been addressed with respect to variables discussed in the previous sections. Increased acceleration at the tibia during Type II landings can be accounted for in the fact that there is a limited range of motion in both the ankle and knee joint at impact. This effectively creates a stiffer leg with less shock absorption. The difference between the ZGS and 1G can also be accounted for by the increased flexion at the knee in simulated microgravity, with 1G knee flexion significantly less ($p<0.001$). Paired t-tests were able to show with an 80% power that all non-significant accelerations were within 6.6 g of one another (Appendix B).

**Figure 4.17:** Filtered Tibial Acceleration During Landing
<table>
<thead>
<tr>
<th></th>
<th>Shank Shock</th>
<th></th>
<th>Foot Shock</th>
<th></th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Tibial Acceleration (g)</td>
<td></td>
<td>Calcaneal Acceleration (g)</td>
<td></td>
</tr>
<tr>
<td>L1</td>
<td>30.47±16.99</td>
<td></td>
<td>28.63±17.89</td>
<td></td>
</tr>
<tr>
<td>L2</td>
<td>47.50±20.74</td>
<td></td>
<td>38.14±19.13</td>
<td></td>
</tr>
<tr>
<td>L3</td>
<td>30.90±19.06</td>
<td></td>
<td>27.49±18.63</td>
<td></td>
</tr>
<tr>
<td>T1</td>
<td>36.40±19.55</td>
<td></td>
<td>30.71±17.92</td>
<td></td>
</tr>
<tr>
<td>T2</td>
<td>34.09±18.70</td>
<td></td>
<td>30.31±17.87</td>
<td></td>
</tr>
<tr>
<td>T3</td>
<td>33.15±19.06</td>
<td></td>
<td>25.51±13.85</td>
<td></td>
</tr>
<tr>
<td>T4</td>
<td>30.19±17.17</td>
<td></td>
<td>26.58±13.71</td>
<td></td>
</tr>
<tr>
<td>T5</td>
<td>49.22±22.56</td>
<td></td>
<td>44.95±23.29</td>
<td></td>
</tr>
</tbody>
</table>

Significance:
- L2>L1,L3; L1=L3
- T5>T1-4, T1≈T2≈T3≈T4
- T5>T1-4, T3>T1,T2; T1≈T2≈T4

Table 4.6: Means and Standard Deviation of Shank and Foot Shocks in the Eight Experimental Conditions
(L - landing type, T - tension level)

The gain/attenuation profile of the tibia to the impacting force can be seen in Figure 4.18. The negative magnitude indicates that the frequency content of the acceleration signal is lower than the content in the force signal and based on the shape of the curve in Figure 4.18, the tibial shock and the force are both more powerful at the lower frequencies. The gain/attenuation profile reveals that the shock was better attenuated in the tibia at the higher frequencies and remained constant after reaching the maximum at approximately 10 Hz.

![Gain/Attenuation of the Tibia to the Impacting Force](image)

Figure 4.18: Gain/Attenuation of the Tibia to the Impacting Force
Calcaneal accelerations measured during jumping appear similar to those of the tibial accelerations, rising quickly to the peak soon after landings often followed by a series of oscillations (Figure 4.19). The oscillations occur most likely due to the rebound of the foot after impact or any extraneous movement. Peak magnitude of the acceleration ranged from 25.51 to 44.95 g with the accelerations experienced during Type I landings significantly higher than the toe-heel type landings (Table 4.6). Additionally, calcaneal acceleration was significantly higher in 1G than any of the ZGS conditions, although several of the conditions within the simulator were statistically different (Tension 3 significantly greater than either Tension 1 or 2, p<0.04). Attenuation of the calcaneal acceleration (Figure 4.20) is greater at the higher frequencies remaining almost constant after 10 Hz. Similar trends were found in the tibial and calcaneal gain/acceleration profile suggesting that the heel pad is the primary structure for absorbing shock in the lower extremities with the ankle joint serving a less role.

![Figure 4.19: Filtered Calcaneal Acceleration During Jumping](image)

![Figure 4.20: Gain/Attenuation of the Calcaneal Acceleration Signal and Impacting Force](image)
4.3 Calcaneal Bone Strain

Strain measurements were recorded during jumping for four of the twelve subjects who participated in the study. Strain was recorded during the entire jump cycle but the strain occurring during the landing phase of the jump was of greatest interest in the current study. Maximum compressive strain during landing as well as the corresponding strain rate in both simulated microgravity and 1G were analyzed in order to develop an understanding of the response to the calcaneal bone during exercise and to determine the feasibility of using jumping as a mean to combat space flight-induced osteoporosis.

The new and unique technique of measuring the calcaneal bone strain necessitated intensive insight regarding the factors which would effect the measurement. Every effort was made to adjust the strain readings in order to report accurate values which truly reflected the actual compression in the bone. Because the Capacitec transducer which was used to measure the bone strain was mounted on two external pins, a major concern was the contribution of the contribution of the mass of the pins, sensor, base and target on the overall magnitude of the measured value. Castigliano's Theorem (Appendix F) was used to determine the deflection of the pins due to the Capacitec instrumentation and the mass of the pins themselves. Because the mass on each pin was different (due to the difference in the mass of the target, mounted on one pin, and the base and sensor mounted on the other pin), each pin deflected a different amount. Also, the position of the base and sensor, on either the top or bottom pin, will altering the overall deflection measured by the transducer.

The length and mass of the pins, the mass of the base, target and transducer and the position of this instrumentation was measured for Subject 4 and Castigliano's Theorem applied to determine the deflection of both pins due to the acceleration at landing (Appendix F, Table 4.7). Because the base and sensor are of greater mass than the target and were mounted on the lower pin for the particular subject studied, the distance between the pins will increase at impact and displacement is measured as the base pin deflection minus the target pin deflection. Strain is calculated by converting this displacement to strain by dividing by the original distance between the pins. The resultant magnitude of
the pin deflection strain (Table 4.7) was determined to be almost two magnitudes less than the measured strain. The pin deflection, therefore, had a negligible effect on the overall measured strain and was not taken into consideration for any of the strain measurements.

<table>
<thead>
<tr>
<th>Target Pin Deflection (m)</th>
<th>Base Pin Deflection (m)</th>
<th>Displacement Between Pins (m)</th>
<th>Pin Strain Due to Impact</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.8x10^-6</td>
<td>2.11x10^-6</td>
<td>3.11x10^-7</td>
<td>2.34x10^-5</td>
</tr>
</tbody>
</table>

Table 4.7: Pin Deflection and Strain by Castigliano's Theorem

The other adjustment necessary in the strain data was the subtraction of the zero value. The zero value consisted of the value measured by the transducer while the subject was non-weight bearing. This value was not recorded properly in the first of the two subjects instrumented with the strain gauges (Subjects 2 and 4). Therefore, strain values calculated using the recorded zero values from the last subjects (Subjects 8 and 9) were used to adjust the strain for Subject 2 and 4. Because of the concern of a shift in the base voltage reading of the transducer, the zero value was plotted as a function of the trial number (Figure 4.21). Both subjects exhibited a discontinuity in the zero reading after the first several trial but remains relatively constant thereafter which provides confidence in the fact that there was no drift in the baseline. The discontinuity which is present in the zero reading at around trial 10 can most likely be attributed to reattachment of the grounding cable.

![Figure 4.21: Recorded Zero Strain Values Versus Trial Number for Subject 9](image-url)
During strain data processing, the zero value subtracted from each data point was determined by two different methods. First, the measured zero value was used. Secondly, the zero value was estimated from the mean strain of the ten points during mid flight. The difference in the calculated strain zero value and the measured zero strain value was determined for the two subjects with accurate zero readings. The average difference for the compressive strain across the two subjects was 0.00143 ± 0.002091. The calculated strain values of the first two subjects were adjusted by subtracting this value from each data point. Therefore, in the final analysis, the strain data for Subjects 8 and 9 were determined using the measured zero value and strains reported for Subject 2 and 4 were determined by using the calculated zero value during mid-flight with a further adjustment made based on the difference in calculated zero and measured zero strain of Subjects 8 and 9.

4.3.1 Peak Compressive Calcaneal Strains

The strain observed during jumping is illustrated in Figure 4.22. It is important to note that while this pattern was seen in many of the jumps, a large amount of variability both between subjects and within subjects was evident in the strain traces. Several important components of the strain are highlighted in Figure 4.22. The peaks in the strain signal indicate the time at which the landing occurred. Tensile strain was observed in the countermovement phase of the jump (prior to 0.5 seconds) while compressive strains were seen during flight. A surprising finding in the strain data was the presence of substantial tensile strain during landing. Initially, it was believed that the high impact force as the subject contacts the supporting surface would exclusively cause a compression in the bone. At times, tensile strain was greater than compressive strain. There is also a noticeable tensile strain during the push-off phase of the countermovement. This was evident in all of the subjects and at times was greater than the strain during landing. Additionally, the strain did not return to a zero value during the flight phase. These events can most likely be attributed to the muscle action acting on the bone.
Figure 4.22: Representative Calcaneal Strain During Jumping.

Upon further inspection of the landing strain, a large amount of decaying oscillations are present after the initial peaks (Figure 4.23). Since the Capacitec transducer has a very high frequency response it was able to capture these oscillations which were probably due to the vibration of the pins during impact as well as the filtering effects used during processing. Since peaks strains always occurred before the subsequent vibrations, the data for the current investigation were not affected.

Figure 4.23: Calcaneal Strain Versus Time During Landing

Calcaneal strains during landing ranged from 0.38 to 0.57% (Figure 4.24). As expected, the flat-footed landing was significantly greater than either of the toe-heel
landings ($p < 0.0001$). There was no significant difference in compressive strains for any of the tension levels (Figure 4.25), once again emphasizing two important points. First, the data show that the strain elicited during 1G can be obtained in simulated microgravity. Additionally, the necessity of using high tension in the zero gravity simulator (which can be uncomfortable for subjects and inhibit their performance) is eliminated since there is no difference in the data. Paired t-tests showed that the values of strain are within 0.2% of one another for each tension level (Appendix B).

![Figure 4.24: Peak Compressive Calcaneal Strains During Landing](image1)

Landing I: Two Feet Toe-Heel; Landing II: Two Feet Flat-Footed; Landing III: One Foot Toe-Heel

![Figure 4.25: Peak Compressive Calcaneal Strains at Five Tension Levels](image2)

Tension 1: 45%BW; Tension 2: 60%BW; Tension 3: 75%BW; Tension 4: 100%BW (± 1 Standard Deviation)

BW - Body Weight
In order to determine if the magnitude of strain recorded during jumping was significantly different than those during low impact activities, the strain was recorded during several walking trials for two of the subjects (Table 4.8). For Subject 8, the strains are at least one magnitude less in walking than those recorded during jumping. These data reveal that the technique used to measure calcaneal strain is capable of discerning different magnitudes of strain. As anticipated, walking results for Subject 8 show that the strain is lower during low impact activities such as walking where the forces reaches levels just over 100% body weight than high impacting jumping exercises where force can exceeded 4 time body weight. This information suggests that the Capacitec instrumentation could differentiate magnitudes of strain in the calcaneus.

<table>
<thead>
<tr>
<th>Subject 8</th>
<th>Maximum Force (N)</th>
<th>Peak Compressive Strain (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>515</td>
<td>0.011</td>
<td></td>
</tr>
<tr>
<td>515</td>
<td>0.0705</td>
<td></td>
</tr>
<tr>
<td>538</td>
<td>0.046</td>
<td></td>
</tr>
<tr>
<td>529</td>
<td>0.06</td>
<td></td>
</tr>
</tbody>
</table>

Table 4.8: Peak Strains at Heel Strike During Walking for One Subject

4.3.2 Compressive Strain Rates

Strain rates during landings were calculated by regressing over a series of points following landing. Because the strain data were so variable, the regression window was chosen manually in the LabVIEW program. This window was defined to encompass the first two peaks in the strain data. Figure 4.26 illustrates the portion of the strain profile used to compute the strain rate (complete strain profiles represented in Figures 4.22 and 4.23). Once the regression window was selected, the midpoint between the maximum and minimum was determined and the regression performed over the five points on either side of this central value.
Strain rates ranged between 2.5 to 126% strain per second in the zero gravity simulator and in 1G. There were significant interactions in the strain rate data with the combination of landing Type II and tension 1G being significantly different from several other conditions (Figure 4.27). A most notable feature of the strain was the fact average peak value was negative in many conditions. This implies that the first peak was a tensile peak instead of a positive peak as in the ZGS tension. It may suggest an alteration in the muscle activation upon landing; however, no significant trend was evident for the conditions. This finding also emphasizes the fact that tensile strain can be a critical factor in the mechanical environment in the calcaneal bone.
The fact that the strain rate had no significant differences for any condition involving landing Type III and only one condition for the Type I landing (both toe-heel landings) may imply that the muscular action and coordination during landing plays a large role in the rate at which the bone is strained. The toe-heel landing required a larger amount of control for the subjects - especially in the zero gravity simulator where a concentrated effort was necessary to make heel contact with the supporting surface. The postural control necessary to execute these landing could have effected the forces acting on the bone through the muscles tendons and ligaments which attach to the calcaneus diminishing the overall rate at which the bone was strained.

4.4 The Relationship Between Force and Strain

A major goal of this research was to develop a relationship between the external ground reaction force and the resultant internal bone strain. Since it was unclear which of the force and strain parameters would be best suited for this goal, the peak force, loading rate, peak strain and strain rate were all considered. Correlations were performed to determine if a relationship between the force and strain was present. Correlations were broken down for landing type (Figure 4.28) and tension level (Figure 4.29) and include the relationship between strain and force, strain rate and force, strain and loading rate and strain rate and loading rate.

![Figure 4.28: Correlations Between Strain and Force for Three Landing Types. Landing I: Two Feet Toe-Heel; Landing II: Two Feet Flat-Footed; Landing III: One Foot Toe-Heel](image-url)
No relationship was evident between any of the force and strain parameters for the tension and landing conditions. Correlations between force and strain tended to be higher in 1G compared to the zero gravity simulator; however, even these values were below 0.55. Correlations performed for subjects individually yielded poor results, although they were better than the grouped results in Figures 4.28 and 4.29. For the four subjects correlations between force and strain were -0.633, -0.322, -0.407 and -0.087 for Subject 2, 4, 8 and 9 respectively (Figure 4.30). The lack of any substantial correlation could be due to the large amount of variability in the strain data both between and within subjects. The trend that does appear in the correlation data is that the strain and force parameters are negatively correlated in all but the Type I landing conditions. Additionally, the 1G data and the Type II landing exhibit correlations which are all negative.
Figure 4.30: Force Strain Correlations by Subject

F/S: Force-Strain; F/SR: Force-Strain Rate;
LR/SR: Loading Rate-Strain Rate; LR/S: Loading Rate-Strain

Figure 4.31 illustrates the interaction between the strain and the force with increasing trial number. A large scatter is present and no trend seems to emerge from the data - even within the individual subjects. A similar trend is found in the interactions between other strain and force parameters (loading rate, strain rate) which is expected due to the high correlations between the peak magnitude and the time rate of change of each parameter. With such great inconsistency, it is unlikely that any relationship could be developed. The source of the variability must be determined. If the variability is due in part to the experimental set-up, then future experiments can address this issue and correct it. However, the wide range may be due to more physiological conditions which have not been measured in the current study. These could include the internal forces (bone-bone, muscles, tendons, ligaments) acting on the calcaneus and controlling the strain environment in the bone. It is entirely possible that these internal forces differ greatly from subject to subject depending on the different jumping mechanics and the recruitment of individual muscles. Additionally, the shape of the bone can contribute to the variability due to the bending which occurs in numerous directions and may not be consistent from subject to subject.
4.5 Cadaver Results

One of the objectives of this study was to determine the relationship between the cadaveric data gathered during drop test experiments and the \textit{in vivo} data in 1G and simulated microgravity. A solid relationship between these data sets would prove extremely useful for future experiments and provide a great deal of insight into the response of the human body to high impact forces without the invasive measures necessary during \textit{in vivo} experimentation. A good example of this idea would be the use of the tibial strain measured in the cadavers. Once a relationship is generated between tibial strain and ground reaction forces in the drop test data, this relationship can be applied to the experimental data to estimate the anticipated strain in the tibia during jumping activities in the zero gravity simulator, a variable which was not measured. The comparison of the cadaver data and experimental data can also, indirectly, give an estimate with regard to the extent to which the muscles and other internal forces are controlling the bone strain. Lastly, this comparison is a means of validating the cadaveric data and realizing its physiological significance. All data was averaged across all trials in both data sets for ease of comparison. Therefore the effect of landing type, tension level or drop height was not be considered for this analysis.

The problem, however with both sets of data (\textit{in vivo} and cadaveric) was the large amount of variability present both between and within subjects or specimens. As
mentioned previously, there was no straightforward relationship between the \textit{in vivo} calcaneal strain data and the external force (Section 4.4). Similarly, no relationship was found for the cadaver strain and force data (highest correlation was between peak force and peak compressive strain - correlation = 0.185 when all data were grouped together). This finding presents a difficult situation with regard to comparing the two data sets and therefore a more qualitative approach must be taken. By specimen, however, there were significant relationships between the force and strain data (Figure 4.32). This suggests that some type of normalization must occur (either by body weight or calcaneal bone mineral density) if an overall relationship is to be found.

\textbf{Figure 4.32:} Correlation Between Force and Strain for Each Cadaveric Specimen

Specimen Number on x Axis
Tibial and calcaneal accelerations were measured on the cadaveric specimens and compared to the *in vivo* data (Figure 4.33). There is no apparent difference between these two data sets which is a critical finding if data from drop test experiments is to be used to predict physiological occurrences of such factors as tibial bone strain (D'Andrea *et al*., 1997). Additionally, a significant correlation between the acceleration data and the force data was found in the cadaveric specimens (mean value across specimens = 0.76) and a weaker correlation between calcaneal acceleration and strain (mean value across specimens = -0.46) suggesting that acceleration data may be a viable means to assess the effects of impact forces on skeletal loading.

![Figure 4.33: Average Acceleration Measurements in Cadaver and In Vivo Data](image)

Average compressive calcaneal strains measured in the cadaveric feet and *in vivo* differed by almost one order of magnitude (0.0407% strain in the cadavers; 0.271% strain *in vivo*; Figure 4.34). Because the average peak ground reaction force was lower in the drop test experiments (Table 4.9), it was expected that the compressive strain would also be lower. However, when the only cadaveric data points considered were those in which the peak force was greater than 1000 N (a magnitude which 95% of the *in vivo* force data fell above), the strains were still significantly lower in the cadaveric specimens (0.0464% vs. 0.271%). Tensile strain was also lower for the drop test but not as severe a difference (0.212% strain in the cadavers; 0.39% *in vivo*).
Figure 4.34: Cadaveric and *In Vivo* Averaged Strain Measurements

Loading rates appeared higher in the human subject studies (Table 4.9). The increased force and loading rate tend to favor the higher strain in the *in vivo* data and is mainly due to the large difference in overall weight. For more accurate comparisons, future studies should consider an adequate normalization variable for both the force and the strain common to both the cadaveric specimens and the human subjects. Another interesting finding in the cadaver data was the magnitudes of the strain rate data. Although the strains in the cadavers were lower, the strain rates were higher than the *in vivo* data by a factor of about 50%. Certainly, the fact that there was no active musculature during the drop tests played a large role in determining the overall magnitude of this variable possibly increasing the resultant value. A more likely explanation, however, is the fact that the strain rates were not calculated in the same manner for the two data sets which led to these discrepancies.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Cadaver</th>
<th>In Vivo</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force (N)</td>
<td>1518±603</td>
<td>2303±727</td>
</tr>
<tr>
<td>Load Rate (kN/s)</td>
<td>198±106</td>
<td>305±169</td>
</tr>
<tr>
<td>Strain Rate (%/s)</td>
<td>23±49</td>
<td>16.2±149</td>
</tr>
</tbody>
</table>

Table 4.9: Force and Strain Parameters in the Cadaver Drop Tests and *In Vivo*
4.6 Results from the Rheological Model

The rheological model was tested at various velocities and with several different masses assigned to the accelerating block. The resultant input force from the block, force in the Hookean spring of the bone, the deformation of the fat pad and the strain in the spring were examined (Table 4.10). From the data it is evident that both the strain and the force increase with the mass and velocity. This proportional relationship yields very high correlations between the bone strain and the force, something that was expected in the experimental data ($R^2 = 0.99$). The force deformation curve shows this linear relationship (Figure 4.35).

<table>
<thead>
<tr>
<th>Condition</th>
<th>Mass (kg)</th>
<th>Velocity (cm/s)</th>
<th>Spring F (N)</th>
<th>Bone Strain (%)</th>
<th>Fat Disp (mm)</th>
<th>Block F (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>M1V1</td>
<td>2.5</td>
<td>25</td>
<td>197</td>
<td>0.018</td>
<td>1.6</td>
<td>198</td>
</tr>
<tr>
<td>M1V2</td>
<td>2.5</td>
<td>50</td>
<td>573</td>
<td>0.052</td>
<td>2.2</td>
<td>574</td>
</tr>
<tr>
<td>M1V3</td>
<td>2.5</td>
<td>75</td>
<td>1057</td>
<td>0.095</td>
<td>2.6</td>
<td>1058</td>
</tr>
<tr>
<td>M1V4</td>
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<td>100</td>
<td>1630</td>
<td>0.145</td>
<td>2.9</td>
<td>1630</td>
</tr>
<tr>
<td>M2V1</td>
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<td>25</td>
<td>335</td>
<td>0.031</td>
<td>1.8</td>
<td>336</td>
</tr>
<tr>
<td>M2V2</td>
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<td>967</td>
<td>0.089</td>
<td>2.5</td>
<td>968</td>
</tr>
<tr>
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<td>1790</td>
<td>0.162</td>
<td>3</td>
<td>1792</td>
</tr>
<tr>
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<td>0.247</td>
<td>3.4</td>
<td>2753</td>
</tr>
<tr>
<td>M3V1</td>
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<td>25</td>
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<td>0.042</td>
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<td>460</td>
</tr>
<tr>
<td>M3V2</td>
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<td>50</td>
<td>1313</td>
<td>0.121</td>
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<td>1322</td>
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<tr>
<td>M3V3</td>
<td>7.5</td>
<td>75</td>
<td>2440</td>
<td>0.222</td>
<td>3.3</td>
<td>2439</td>
</tr>
<tr>
<td>M3V4</td>
<td>7.5</td>
<td>100</td>
<td>3716</td>
<td>0.336</td>
<td>3.7</td>
<td>3720</td>
</tr>
<tr>
<td>M4V1</td>
<td>10</td>
<td>25</td>
<td>566</td>
<td>0.052</td>
<td>2.2</td>
<td>565</td>
</tr>
<tr>
<td>M4V2</td>
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<td>1640</td>
<td>0.15</td>
<td>2.95</td>
<td>1641</td>
</tr>
<tr>
<td>M4V3</td>
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<td>3.5</td>
<td>2996</td>
</tr>
<tr>
<td>M4V4</td>
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<td>100</td>
<td>4584</td>
<td>0.417</td>
<td>3.9</td>
<td>4589</td>
</tr>
</tbody>
</table>

Table 4.10: Rheological Model Results
(Twelve Tested Conditions)
The ranges of the strains and forces are well within those which are found in the experimental data which indicates that the model is capable of predicting the physiological events which occur in the bone when subjected to high impact force (Figure 4.36). The predicted bone strain from the model appears significantly less than that measured *in vivo*. However, it is important to note that the magnitude of the strain can be significantly altered by changing the initial mass and velocity provided the peak force stays within the given range measured during experimental data collection.

Strain rate was measured in the model as the slope in the most linear portion of the strain versus time curve. Strain rate in the model was within the range of the values measured *in vivo* (Figure 4.37). The model, however, tended to over predict the strain rate which could be due in part to the method of calculating the variable. The surprising results is that a clear relationship was found for the force and the bone strain in the model and not in the experimental *in vivo* data. This suggests the potential for developing a similar relationship in the *in vivo* data. Further manipulation of the experimental data and determination of any artifacts which are present will be necessary to achieve this goal.
The effect of the heel pad was seen by analyzing the deformation in the bone in the calcaneal trabecular bone model alone and the final hindfoot model which incorporated both the bone and the viscoelastic fat pad. Applying an equivalent force to the bone only model and the hindfoot model, the strain was reduced by up to as much as 70% when the heel pad was attached showing the significant role of the heel pad structure. The deformation in the fat pad also matched well with those values measured in the literature. De Clercq et al. (1994) found that the fat pad was deformed a maximum of 60% during running studies. In the model, the strain in the fat pad reached a maximum of 40% for the conditions under which the deformation was measured (Table 4.10; Figure 4.38). This
value is lower than the obtained by De Clercq et al. (1994) but the higher forces which act over a shorter period of time in jumping may effect the damping properties of the heel pad causing it to become less compressible.

![Fat Pad Strain in Rheological Model](image)

**Figure 4.38:** Fat Pad Strain in Rheological Model

One of the interesting facts that was brought out in the model was the relationship between the stiffness of the bone. It can be argued in the same statement that the model should be stiffer and should become less stiff in the same argument. Keeping the input force the same, if the damping of the bone is decreased by adjusting the coefficient of dashpot, the overall stiffness calculated by the slope of the force-deformation curve will become greater (Em). However, this will cause the strain rate to increase, which in turn causes the stiffness calculated by the equations of Carter and Sprengle (1982), Ec, to decrease. This was a critical factor in determining the value of the damper coefficient (which was chosen so that the two stiffness were close in value).

The possible causes of this phenomenon is related to the relationship between the viscosity (damping) and the apparent density in bone. In the theoretical development of the model to determine $\eta$, the density remained constant while the damping was altered over a wide range of values. This may not be physiologically possible in bone, however. For a given density, there may be only a certain small range over which the viscosity may lie.
4.7 Summary

This chapter reviewed the biomechanical data related to jumping in simulated microgravity and in 1G. It was able to show the similarities and differences between these two conditions as well as the ability of different landings to yield a wide response with regard to these jumping parameters. The results of a new technique for measuring *in vivo* trabecular bone strain were reviewed in order to assess the feasibility of this new method. Relationships between the internal bone strains and the external ground reaction forces were not present in the data; however, the cadaveric results and the rheological model showed promising trends in approaching the solution to this hypothesis.
CHAPTER 5

DISCUSSION

The overall purpose of the current investigation was to ascertain the feasibility of jumping exercises as a means of preventing space flight induced osteoporosis, a major physiological adaptation which occurs during microgravity. The underlying hypothesis in achieving this goal was the idea that high impact forces imparted to the body are a major controlling factor in skeletal homeostasis. While other investigators have determined the level of strain required to maintain the cortical integrity (Rubin and Lanyon, 1987), the relationship between the external forces acting on the body and the resultant internal response has yet to be ascertained. By recording a wide variety of measurements during jumping, including calcaneal bone strain, acceleration, joint angles and external ground reaction forces, this investigation provided a unique opportunity to examine this relationship. Not only were these parameters recorded in a 1G environment but also in simulated microgravity in order to give credibility to the ability of these exercises to be effective in space.

5.1 Impact Forces

The two major components existing in this study were the presence of high impact forces and the ability to measure internal bone strain. Jumping exercises were chosen because of their known ability to impart high forces to the body (Bobbert et al., 1987b; Dufek and Bates, 1990; Dufek and Bates, 1991; Mizrahi and Susak, 1982). In a review of the recent literature, Dufek and Bates (1991) reported up to 15.1 times body weight during landing in certain athletic activities. Mizrahi and Susak found force peaking at 4.26
times body weight while landing flat-footed from a free fall. The variability in the peak magnitude, however, was great. The landing force is dependent on many variables - impact velocity, jump height, landing technique, stiffness of the supporting surface and the joint motion. The effects of several of these factors were measured in the current investigation and their effect on the peak force in landing determined.

The fact that the jumping exercises can impart high impact forces alone does not necessitate that there is any significant effect on the bone. However, a large body of data has been amassed regarding the ability of high impact exercise to have a positive effect on the bone mineral density in the lower extremities (Chilibeck et al., 1995; Dyson et al., 1997; Forwood and Burr, 1993; Heinonen et al., 1996; Snow-Harter et al., 1992). This has created a strong foundation supporting the idea that the mechanical stimuli from the external force is responsible for the internal stimuli to enhance bone formation. Therefore, with the high impact forces associated with jumping, it can be expected that this type of activity will have a definite effect on the bone mass - especially in the lower extremities, and particularly in the foot, where the force has not been extensively attenuated by the soft tissue.

In the current investigation, the average peak forces were between 1902 and 2631 N. The flat-footed landing imparted the higher force than the two toe-heel landings, congruent with the findings of Mizrahi and Susak (1982) who found the toe-heel contact pattern during landing to minimize the lower extremity loads compared to the flat-foot landing. The second issue concerning the ground reaction force is the ability of achieve magnitudes in simulated microgravity similar to 1G levels. In a study analyzing peak force and loading rates in exercise most commonly used during space travel, Cavanagh et al. (1992) contend that although the exercises such as running appear to possess high loads and loading rates, these values may not be present in microgravity exercises. This may be due in part to the mechanics used by astronauts during such activities as running in space (Convertino, 1990). The current results support the fact that it is difficult to realize peak force and loading rate values in microgravity that mimic those in 1G. Both the force and the loading rate were significantly higher in 1G compared to the simulated microgravity.
conditions. This may be attributed in part to the experimental set-up in the simulator and the inclination of the subject to bend their knees. Further studies in true microgravity are necessary not only to validate the effectiveness of the zero gravity simulator but to determine the magnitude of the diminished jumping force in microgravity.

The values obtained for the peak load compare well to the range reported for jumping in simulated microgravity (Vrijkotte, 1991). However, discrepancies exist between the values of the loading rate. In the current investigation, values ranged between 221 and 426 kN/s, almost three times greater than those reported in the previous study. This is due to the manner in which they were calculated. Vrijkotte (1991) determined the loading rate by performing a regression over the entire period from initial impact to peak force. While the higher values in the current study were obtained from the slope of the force history profile near the peak. The load and loading rates during jumping in simulated gravity appear to be substantially higher than those reported for running and walking in a zero gravity simulator (Davis, 1991; McCrory, 1997). This further substantiates the concept that jumping in microgravity has the potential for imparting forces which may be adequate to stimulate bone deposition.

5.2 Factors Affecting Peak Force Magnitude

5.2.1 Jump Height, Impact Velocity and Flight Time

Parameters which significantly affected the overall force and loading rate were the impact velocity, jump height, flight time and the joint motion. There was no significant effect on the magnitude of the peak force at landing due to the tension levels in the zero gravity simulator. This was due in a large part to the height of the jump which decreased with increasing tension. As shown by Bobbert et al. (1987), the peak vertical ground reaction force was significantly increased during drop landings from 20 to 60 cm in 1G. As tension increased in the simulator the subjects had the ability to exert more force at landing. However because the tension was higher the subject could not produce enough push-off force to counteract the gravity replacement system at the higher levels and achieve jump heights reached in the lower tensions. 1G levels for jump height are
comparable to the values at the 45% tension level. These finding suggests that the higher tensions are not necessary in the zero gravity simulator, a significant finding considering the discomfort to the subjects at the higher tensions. It also has implications for the force in the tethering system pulling astronauts towards the exercise equipment aboard the space shuttle when considering the impact forces that are generated. In both cases, a lower tension would decrease discomfort and restriction and help subjects maintain more "normal" mechanics during activity. The impact velocity, jump height and flight time are all highly correlated and showed a significant increase in value in 1G. Additionally, the impact velocity was significantly higher at 45% body weight than at the other ZGS tension levels in the Type I and Type II landings. These factors contribute to the acceleration levels experienced at impact and therefore affect the landing force.

5.2.2 Joint Kinematics

Significant variables which must be considered when reviewing the magnitude of the peak force are the kinematics of both the ankle and knee. The ability to flex the joints allows the magnitude of the incoming shock transient to be attenuated. Several researchers have investigated the effect of the knee and ankle on this shock absorption (Gross and Nelson, 1988; McMahon et al., 1987; Radin et al., 1972). In the current study, the knee and ankle sagittal plane angles were determined during the jumping exercises in both simulated microgravity and 1G. While the magnitude of the knee angle at the time of landing was thought to be the critical parameter in the force attenuation, the ANOVA results showed significantly higher knee flexion angles associated with the greater impact forces experienced during the flat-footed landings. The more telling variable was the range of motion of the joint from impact to peak angle. The lower ranges for knee flexion were found in the Type II landings helping to explain the significantly greater impact forces.

The ankle dorsiflexion angle during impact occurs in the closed kinetic chain. The peak angle is higher for the Type I landing and can be attributed to postural control. In the toe-heel landings, the subjects were instructed to make sure the heel made contact with the supporting surface. By moving the tibia over the grounded foot (thus creating a
dorsiflexed angle), the heel is more likely to contact the ground. Considering this phenomenon, it would be expected that Type I and Type III landings would be similar. However, a significant difference was found in this case with Type I landings having greater dorsiflexion angles. This can only be explained by the fact that the subjects had difficulty landing on one foot and required greater balance control. This was most likely achieved by co-contraction at the joints, thus limiting the range of motion. Another surprising result was the fact that the ankle angle was significantly greater in 1G than any of the ZGS conditions. The fact that there is a propensity to bend the knees in the simulator would seem to indicate that the ankle angle would be less in 1G. This, however, is not the case.

5.3 Acceleration  

The transmission, absorption and attenuation of the shock transient after impact is an important component of the bone physiology. This shock wave can be measured with accelerometers mounted on the calcaneus and tibia. The calcaneal accelerations indicate the shock absorbing effects of the heel pad while the acceleration measurements in the tibia reveal the attenuation due to the ankle angle. Peak accelerations were greatest for the Type II landings (46 g) which is a reflection of the increased impact force in this condition. Both the tibial and calcaneal acceleration were greater in the two-footed toe-heel landings compared to Type III one-foot landings. This result is in direct contrast to the force which is significantly greater for Type I landings. Again, a possible explanation is the "stiffer" landing on one foot.

Because the force is continually absorbed as it travels through the body, it is expected that the tibial acceleration readings would be less than that of the calcaneus. However, in the current investigation the tibial strain was higher than the calcaneal strain in most jumps. This is due in part to the fact that the accelerometer was capable of measurement in only one direction. The accelerometer was mounted on the calcaneus so that the values were recorded parallel to the long axis of the body. In the toe-heel landings the heel is not flat on the ground at impact and the resultant acceleration is no longer along the long axis of the body but rather offset by the angle at the ankle. Because
the acceleration is recorded in only one direction, the resultant value cannot be resolved into the vertical direction. Since the tibia is assumed to be vertical at impact this problem should not be present. The fact that the knee moves over the ankle following the landing may, however, make this assumption invalid and introduce errors in the data. Additionally, it has been shown that soft tissue can cause a higher frequency response in the acceleration data. Because the lateral surface of the calcaneus does not present a solid bony prominence as the tibial crest, this phenomenon may be present in the calcaneal data.

Frequency response of the acceleration data was examined by determining the gain/attenuation at both the tibia and the calcaneus. The values were averaged for all subjects and the resultant profiles showed a increased attenuation at the higher frequencies. This is in agreement with impact data reported by Lafourtue and Lake (1995).

In the frequency domain, the power of the signal indicates the body’s capability to alter the input shock. As mentioned previously, the similarities in gain/attenuation suggest the goal of the body is to diminish the shock above all else. Not to be ignored, however, is the close proximity of the two accelerometers in this investigation. Previous studies (Lafourtue and Lake, 1991; Voloshin, 1981; Voloshin and Wosk, 1982) studied the shock transmission but used points on the tibia and head revealing a large attenuation. The differences in the calcaneal and tibial acceleration may not be discernible because of the relatively short distance over which the shock waves are measured.

5.4 Calcaneal Bone Strain

Force and acceleration readings are useful in understanding the magnitudes of the shock transient through the body but the effect of jumping on the calcaneal bone mass cannot be determined without an understanding of the relationship between the external force and the internal bone strain. This necessitates the measurement of the strain during jumping. Previous in vivo investigations have utilized rosette strain gauges mounted directly in the tibia (Burr et al., 1996; Lanyon et al., 1972). The shape and orientation of the calcaneus prohibit the use of this technique. By drilling the two k-wires directly into the calcaneus and, with the use of the Capacitec transducer, the bone strains can be measured in the calcaneus in a less invasive manner. Another exciting component of this
research is the fact that it is the first to document the \textit{in vivo} strains in trabecular bone. The previous studies were limited to cortical bone which is stiffer and with regard to the current study, does not exhibit the loss of bone mineral density in space to the extent of the trabecular bone.

Because the technique used in measuring the calcaneal bone strain had not been used previously, several problems presented themselves early in the investigation. As discussed in the previous section, the vibration of the pins upon impact can cause a resultant deflection due to mass of the pins and strain gauge instrumentation. Using Castigliano’s Theorem, the magnitude of this deflection was small enough to have little or no effect on the resultant strain. There is, however, the problem related to the fact that the pins are in essence two separate systems and can act independently from one another.

The anatomy of the calcaneus is such that the outer surface consists of thin layer of cortical bone which surrounds the inner trabecular bone. Since the pins are drilled into the lateral side of the calcaneus from the surface of the skin, they must first penetrate the cortical bone which can have pronounced effects on the strain measurements. In the first case, the cortical bone, because of its marked increase in stiffness compared to trabecular bone, will limit the strain read by the transducer by restricting the movement of the pins. The pins will not move freely in the trabecular bone and the strain will be underestimated. Conversely, an alteration in the strain may be recorded if the pins separate independently from one another in the "softer" trabeculae upon impact. This can be referred to as the “teeter totter” effect. Exaggerated tensile strain would appear if the ends of the pins in the bone moved closer to one another causing the opposite ends to spread apart. Conversely, if the ends in the bone more away from one another, the compressive strain measured by the Capacitec transducer would be larger. It is unknown if the high impact forces associated with jumping are capable of moving the pins independently and creating either of these situations but the possibility of their existence must be investigated.

The average peak magnitudes of the compressive strain ranged between 0.34 to 0.57% strain with individual strain reaching as high as 1.36%. Three different landing types were utilized in this study to realize a wide spectrum in the strain response. The
compressive strain was significantly higher for the flat-footed landing but no significance
can find with regard to tension level. This was an unexpected result since it was
expected that strain was related to force which was statistically greater in 1G compared to
the ZGS tension levels. On one hand, the fact that there is no difference between 1G and
the simulated microgravity conditions suggests that the strain during jumping obtained on
Earth can be achieved in space. However, the fact that the strain does not follow the same
trend as the force creates problems in developing a relationship between these two
variables.

The magnitude of the compressive strains during jumping showed promise with
regard to the ability of jumping to produce osteogenic stimuli. Rubin and Lanyon (1987)
found that a level of 0.1% strain applied to the cortical bone at least four times a day is
sufficient to maintain skeletal integrity and prevent endosteal resorption. The fact that the
jumping strains were 4 to 5 times greater than this threshold implies that this exercise is
osteogenic in nature. With such a large difference in peak magnitude, it is possible that
even with altered mechanics of jumping in a microgravity environment resulting in
decreased forces, the strain experienced by the bone may still reach this osteogenic level.
Further studies of jumping in true microgravity are necessary to prove this hypothesis.

One factor which must be considered is the fact that the shock transient
experienced by the tibia is diminished when compared to that seen in the calcaneus. This
implies that the force available to strain the tibia is reduced. Therefore, if the osteogenic
level in the tibia cortical bone is 0.1%, the strain in the calcaneal bone must be higher,
especially if it is to be an osteoregulatory stimulus. The fact that the cortical bone is
significantly stiffer than trabecular bone supports this hypothesis.

The fact that the high impact stimulus can be applied a very few times and achieve
the same results (more than 36 cycles per day did not significantly increase bone
deposition in studies by Rubin and Lanyon, 1984) is very appealing for exercise in space.
Time is critical for astronauts traveling in space and the less time they have to spend
exercising, the more time they can devote to their work. With jumping, several trials can
be executed in less than a minute and, if the relationships proposed by Rubin and Lanyon (1984) are not altered in space, astronauts can complete training in no less than four repetitions.

One of the major concerns in using the above osteogenic threshold and cycle number is the fact that they apply to cortical bone while the current investigation has focused on the trabecular bone in the calcaneus. Cortical bone is stiff with little variation in the modulus and strength properties (Keaveney and Hayes, 1993). It is distinguished by an osteonal structure and has low porosity. Trabecular bone, on the other hand, is relatively compliant, very heterogeneous with high porosity and open-cell architecture. These differences give the two types of bone very different and unique mechanical properties. No conclusive evidence has been found showing that the strain environment is similar in both types of bone. The fact that the trabecular bone is filled with the fluid-like marrow may significantly alter its response to external impact. Therefore, the assumption that the data reported by Rubin, Lanyon and colleagues can be used to define the osteogenic threshold in the calcaneus is uncertain.

Further complicating the matter is the fact that the yield strains for trabecular bone have rarely been reported. Bovine trabecular bone from the distal femur has been shown to yield at approximately 0.74% (Turner, 1989) while Linde et al. (1989) reported a 2.02% ultimate strain in the proximal human tibia. Clearly there is a large variation in the failure strain which is compounded by the dependence of the strength on the apparent density. Further studies are needed to define the properties of trabecular bone, especially related to the applied forces.

Since the impact phase of the jump is generally associated with compression, the fact that the calcaneus experienced significant tensile strains during impact was not expected. Several tests were performed to assure that the tensile strains were not an artifact in the data and conclusive evidence was reached to be certain that the bone was in fact going into tension. During jumping exercises, a compressive peak in the strain time history profile was rapidly followed by a tensile peak with a magnitude which, on average, was 1.5 times the compressive peak. Since the strain rate was calculated as the slope
between these compressive and tensile peak, it reflects the order in which these peaks occurred. Average strain rates were positive in all conditions except for Type III landings (-19 %/s) and in 1G (-58 %/s) suggesting involvement of the neuromuscular system for control. The interaction between landing Type II and Tension level 5 (1G) was significantly greater than Type I*Tension 3, Type II*Tension 1, Type II*Tension 2 and Type II*Tension 3 pointing again to the fact that these two conditions (Landing II and Tension 5 - 1G) show the greatest magnitude for the ground reaction forces.

5.5 The Relationship Between Force and Strain

With the present data it was not possible to determine a relationship between the force and strain parameters as initially intended. Strain data were widely scattered and even within subjects no solid relationships were found - even after separating out the individual factors. Peak correlations did not exceed 0.53 with most values below 0.3.

The lack of any strong relationship between force and calcaneal strain is not surprising given the complexity of the bone and the measurement technique. It is speculated that, upon impact, the irregular shape of the calcaneus causes bending to occur about a neutral axis. Bending could occur in any direction with the possibility of the pins being on the tensile and compressive side of bending for any given jump and significantly altering the strain readings. This bending would account for the unexpected presence of the tensile strains during landing. Error is also introduced into the strain signal because the neutral axis is unknown and the placement of the pins with respect to this line most likely varies between the four subjects instrumented with the strain gauge apparatus. It is difficult to assess the effect of this difference in the current investigation and a more quantitative measurement of the neutral axis is necessary as well as more precise placement of the pins.

Additionally, it is hypothesized the muscular role during jumping is significant and could contribute considerably to the level of strain in the calcaneus. Since each subject may have a different motor strategy for executing a jump, the potential for large differences in the overall bone strain exist between subjects. The force in the achilles tendon, which can reach magnitudes of up to 2000 N during jumping (Fukashiro et al., 1990).
1995), must have a definitive role on the strain. This effect, however, was not measured in the current investigation.

One of the major drawbacks of the current technology is that it is only possible to record the strain in a very small area of the calcaneus. This is not uncommon for such in vivo measurements (Lanyon et al., 1972; Burr et al., 1996) but limits the conclusions that can be drawn from the data. More global measurement in the bone would help to eliminate some of the errors introduced by the calcaneal bending and be able to quantify the action of the muscles and ligaments on the bone.

5.6 Cadaver Results

The comparison of the cadaver data to the in vivo data offered little or no new information regarding the force and strain parameters and the relationship between the two. The impetus in performing such a comparison was twofold: to validate the ability of the cadaveric information to match the physiological conditions and to use relationships in the cadaveric data to estimate variables not quantified in the human subjects due to the invasiveness of the measurement. However, the drop test data was extremely variable and the relationship between the strain and the force parameters was only evident in individual specimens. This suggests the need to find a normalizing factor such as bone mineral density in the calcaneus or perhaps body weight. A more straightforward association was expected in the cadaver data due to the absence of confounding factors such as the muscular activity acting on the bone or excessive joint activity but was not found. This once again stresses the complexity of calcaneal bone and raises questions regarding the accuracy of the strain transducer.

One of the factors that can be considered is the relationship between the measured accelerations and the internal bone strains. Acceleration transients represent the force transient as it travels through the body deforming the soft and hard tissue it passes. If a true relationship between the force and strain can be found, it is possible that the acceleration signal may provide a simple, non-invasive means of assessing the bone strain. The fact that the cadaveric data showed a high relationship between peak force and peak measured calcaneal acceleration support this hypothesis. The fact that the correlations
between acceleration and strain were significantly lower does not diminish this proposed relationship but rather speaks to the fact that the strain values were extremely variable and most likely contain artifacts which need to be removed.

5.7 The Rheological Model

The rheological model provided a means to mathematically describe the factors affecting the peak magnitude of the force and can be easily controlled to simulate different conditions. The relationship between the external force and the bone strain calculated was highly correlated showing somewhat of a cause and effect relationship. This relationship, however, was not linear as shown by the fact that a doubling of the input force did not cause a two fold increase in the bone strain. This is due to the viscoelastic nature of the heel pad through which the transient shock wave must travel before affecting the bone. Although this model is descriptive and does not define exact anatomical features, the springs of the heel pad can be loosely defined as the elastic fat pad septa and the dampers as the viscous, specialized fat globules. Although the coefficients of the spring and dampers in the fat pad cannot be verified (as was done with the bone model), the overall response corresponds nicely to that reported in the literature.

The bone model was developed both theoretically and based on the physiological data measured during jumping exercises. Coefficients for springs in the Hookean and Kelvin units could be verified by mathematical derivation and the equations used to determine bone deformation matched the predictions in the ADAMS® software when equal forces were applied.

Overall, the model is a good predictor of the physiological response of the calcaneal bone to high impact forces. The magnitude of forces, heel pad strain, calcaneal bone strains and strain rates fall well within the range measured experimentally. The unique aspect of the model is that it can be altered to simulate the response to loading regimes in many different conditions such as a stiffened heel pad, as seen in diabetes, or osteoporosis, where the density of the bone is significantly decreased.
5.8 Summary

While the overall cause and effect relationship of impact forces applied to the body and the resultant bone strain remains unclear, this investigation was successful in gaining insight into the strain environment in the calcaneus when exposed to high impact forces. The strain measured during jumping in both 1G and simulated microgravity was significantly higher than the osteogenic threshold proposed by Rubin and Lanyon (1987) indicating that these exercises may be well suited for combating bone loss both in space and on Earth. This investigation also provided insight on a new, challenging technique in measuring \emph{in vivo} bone strain -especially in trabecular bone which has not, as of yet, been documented.
CHAPTER 6

CONCLUSIONS AND RECOMMENDATIONS

In conclusion, this investigation set out to determine the efficacy of jumping as a means to combat the bone loss in space. From the results, it can be concluded that jumping is a viable alternative to exercise during space travel. Not only will this activity help to combat space flight-induced osteoporosis, it also will decrease the amount of time spent exercising allowing the astronauts to devote more attention to the scientific work at hand. This study also introduced a new and unique method of measuring \textit{in vivo} trabecular bone strains. This technique is significant in the fact that there is no other published data on trabecular bone strain during jumping or any other activity. Finally, the fact that this investigation used more than one method to assess the response of the lower extremities to high impact loads strengthens arguments made regarding the measured parameters.

Five specific aims were defined and are highlighted below with respect to the findings of this investigation.

6.1 Specific Aims

\textit{Specific Aim 1 and Specific Aim 2:} To measure and compare the biomechanical aspects of jumping in simulated microgravity and 1G. Ground reaction forces, tibial and calcaneal acceleration, calcaneal strains and loading rates were the biomechanical measurements recorded for jumping in simulated microgravity and 1G. The magnitude of the ground reaction force did not change at the different levels of tension in the zero gravity simulator but was significantly
higher in 1G. Loads also varied by the type of landing with flat-footing landing significantly greater than the toe-heel types. Accelerations were significantly greater in 1G but also increased with peak load. Within the zero gravity simulator, no impact parameters (force, acceleration, etc.) were effected by the tension in the gravity replacement system. This is due to the ability to jump higher at the lower tension levels. Therefore, when jumping exercises are considered, there is no added value in increasing the tension in the GRS springs.

**Specific Aim 3:** To establish a relationship between external ground reaction forces and loading rates and internal strain and strain rates. Although the major impetus of this investigation was to determine this internal/external relationship, it was not possible with the current data. The variability was too great in the strain data and there are too many uncertainties present in the strain measurement technique. The data did give promising results relating to the fact that the bone strain does achieve values above the osteogenic threshold defined by Rubin and Lanyon (1987) during jumping exercises in both simulated microgravity and in 1G. Further research is needed to decrease the variability in the strain data and to eliminate any artifact in the data due to instrumentation errors.

**Specific Aim 4:** To compare impact data from cadaveric drop tests and *in vivo* jumping exercises. Cadaver data provided little insight into the mechanisms relating bone strain and external forces due to some of the same problems which have affected the *in vivo* data. Differences were evident in the two data sets (*in vivo* and cadaveric) for peak force, loading rate, compressive and tensile strain and both tibial and calcaneal accelerations. In all cases the magnitude of the variable in the *in vivo* data set was greater than in the cadaver data.
Specific Aim 5: To develop a model which can accurately predict the response of calcaneal bone strain to impact forces.

This research has successfully shown that a rheological model can be derived which will mimic not only the compression of bone but also the effect of the underlying, viscoelastic heel fat pad. In essence three models capable of functioning independently were developed - the heel pad model, the bone model and the hind foot model which is a combination of the two. With a given input force, this descriptive model was able simulate strains which compared favorably to the experimentally measured magnitudes. Additionally, the model produced the best result regarding the relationship between the external impact force and the resultant strain in the bone where these variables were highly correlated.

6.2 Recommendations for Future Research

(1). Future studies are needed to validate the zero gravity simulator by determining if the same response to jumping can be achieved in true microgravity.

(2). Further validate the manner in which the strain was measured by conducting more cadaveric tests to ascertain whether or not any of the effects discussed above are present and to determine a means to eliminate any problems due to experimental set-up.

(3). More information is required to develop a solid relationship between the external force and the internal bone strain. Further in vivo measurements in highly restricted environments may help to eliminate extraneous influences and produce an association between the two variables.

(4). Further investigation on the properties of trabecular bone are necessary in order to ascertain if the strain relationships described by Rubin and Lanyon (1987) are similar to those in cortical bone.

(5). Obtain a better understanding of the shock attenuation through the lower extremity by obtaining more accurate calcaneal acceleration with a triaxial accelerometer.

(6). Further development of the rheological model possibly incorporating the effects of muscular forces on the bone and bone strain.
(7) Determine the relationship between the acceleration and strain data and assess the ability of calcaneal acceleration measurements to accurately predict bone strain.

(8) Using jumping exercises in long-term microgravity studies to determine the possible effect on bone density.
APPENDIX A

CALIBRATION AND LENGTH OF THE LATEX CORD

To calibrate the latex cords, a series of weights were hung from a known length of the cord and the displacement measured (Figure A.1).

![Load vs Strain for Latex Cord](image)

Figure A.1: Load/Strain Characteristics of Doubled Latex Cord

Using linear regression, the following relationship between the two variables (strain and weight) was found:

\[
\text{Cord Strain} = (0.0958) \times \text{Weight} - (0.477) \quad (A.1)
\]

To eliminate the need for several different lengths of cord, rope was hung in series with the latex cords. Therefore, the length of the rope would be adjusted depending on the weight of the subject and the extent to which the latex cords were stretched. Since the distance between the ceiling bracket and the wall mounted force plate is approximately ten feet, the combination of the rope and cord must cover this distance. Because the objective
was to optimize the length of latex cord and to use as little rope as possible, the weight of the heavier subjects will dictate the length of the cord (heavier subjects will stretch the cord more) and the rope length will be determined by the lighter subjects (lighter subjects will require more rope since they will not stretch the cords as much). Considering the size and weight of all subjects in this experiment, the suspension system was designed to accommodate 120-225 pounds.

The weight supported by the latex cord can be approximated by using anthropometric data (Clauser et al., 1969). The weight of the thigh, shank and foot segments were calculated as 10.3%, 4.3% and 1.5% of the total body weight, respectively. Since the latex cords support only the lower extremities, the length of cords attached to the thigh segment and the combined shank and foot segments were calculated and right/left symmetry was assumed. The initial length of the cords ($L_0$) was determined by knowing the strain for a given body segment mass based on Equation A.1 and the final length (or stretched length) which was set at ten feet due to experimental constraints (Equation A.2).

\[
\text{Cord Strain} = \frac{\text{Final Length} - \text{Initial Length}}{\text{Initial Length}} \quad \text{Cord Strain} = \frac{10 - L_0}{L_0} \quad (A.2)
\]

An iterative method was used to optimize the length of the cord for the thigh (Table A.1) and the shank and foot (Table A.2). Calculating the rope and cord lengths for varied initial lengths, it was determined that the thigh cords should be 3.5 feet long and the shank/foot cord 5.5 feet. Above these values, the system exceeded the ten foot constraint. The length of the rope used for each segment was determined as the length of rope required for the lightest weight subject at the corresponding latex cord length. For the thigh and shank segments, rope length totaled 4 and 3.5 feet, respectively.
### Table A.1: Thigh Segment Latex Cord Length

<table>
<thead>
<tr>
<th>BW (lbs)</th>
<th>Thigh (lbs)</th>
<th>Strain</th>
<th>Stretched Cord L (ft)</th>
<th>Rope L (ft)</th>
<th>Stretched Cord L (ft)</th>
<th>Rope L (ft)</th>
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### Table A.2: Foot/Shank Segment Latex Cord Length

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<th>BW (lbs)</th>
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<th>Stretched Cord L (ft)</th>
<th>Rope L (ft)</th>
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APPENDIX B

Paired T-Test Results

Paired t-tests were performed using the nQuery Advisor 2.0 software (Statistical Solutions, Boston, MA) in order to determine the magnitude of the similarity in variables with an 80% power. The analysis of variance (ANOVA) determined the conditions in which there was no significant difference each of the measured variables. The ANOVA also supplied the difference in the least squares means (as well as the standard error of the difference. In each set of non-significant variables, the greatest difference between conditions was determined. The greatest difference was used in the analysis because it was decided that this would yield the "worst case scenario" for the maximum difference between the non-significant values. For example, no statistical difference was found for the maximum force in any of the four zero gravity tension levels. For the force the greatest difference was 46.13 N (Table B.1) occurring between Tension 1 (45% body weight) and Tension 4 (100% body weight). The standard deviation was estimated from the standard error of the difference between Tension 1 and Tension 4 (SE=69.73, SD=241.55).

Using nQuery, a paired t-test of the means was performed on the comparison of the two conditions with the greatest difference. The test significance level was set at 0.05 for a two sided test and the power set at 80%. The number of subject was inputted to the spreadsheet along with the estimated standard deviation. From this information, the program yielded the mean difference mean difference in each set of non-significant variables (Tables B.1 and B.2). With this information, the range over which the non-significant variable were the same could be reported. In general, the mean difference was approximately 10-20% of the least squares mean.

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### Table B.1: Mean Difference Results for Non-Significant Occurrences in Force and Acceleration Data

<table>
<thead>
<tr>
<th></th>
<th>Max Force (N)</th>
<th>Loading Rate (N/sec)</th>
<th>Max Tibial Acc (g)</th>
<th>Max Calc Acc (g)</th>
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<td>Significance level</td>
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<td>1 or 2 sided?</td>
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<td><strong>59451</strong></td>
<td><strong>6.614</strong></td>
<td><strong>6.298</strong></td>
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<td>Standard deviation of differences</td>
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<td>7.05</td>
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### Table B.2: Mean Difference Results for Non-Significant Occurrences in Strain and Angle Data

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<td><strong>Mean difference</strong></td>
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<td>Standard deviation of differences</td>
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APPENDIX C

Health History Form

Name: ____________________________________________________________
Address: __________________________________________________________
__________________________________________________________
__________________________________________________________
Phone: (___) ________ - ________ (Home)
(____) ________ - ________ (Work)
Social Security Number: ________ - ________ - ________
Date of Birth: ___/___/___
Age: ________
Height: ________ in ________ cm
Weight: ________ lb. ________ kg

1. Are you currently taking any medication on a regular basis? (Please specify)

2. Do you have any history of broken bones, surgery or injury to the lower extremities? If yes, please explain.
3. Do you have any problems with balance? If yes, please explain.

4. Please respond to the following regarding your health history. If you answer yes, specify.

**Cardiovascular/Respiratory History**

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<thead>
<tr>
<th>Condition</th>
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<tr>
<td>Heart disease, past or present</td>
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<td>No</td>
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<td>Heart murmur</td>
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<td>No</td>
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<tr>
<td>High blood pressure</td>
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<td>No</td>
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<tr>
<td>Occasional chest pain</td>
<td>Yes</td>
<td>No</td>
</tr>
<tr>
<td>Shortness of breath</td>
<td>Yes</td>
<td>No</td>
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<tr>
<td>Asthma or allergies</td>
<td>Yes</td>
<td>No</td>
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<tr>
<td>Fainting</td>
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Explain: ______________________________________________________

________________________________________________________________

**Musculoskeletal History**

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<td>Muscle injuries, past or present</td>
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<td>No</td>
</tr>
<tr>
<td>Muscle pain during exercise</td>
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<tr>
<td>Arthritis</td>
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<tr>
<td>Joint injuries, past or present</td>
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Explain: ______________________________________________________

________________________________________________________________

**General History**

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<tr>
<td>Gastrointestinal problems</td>
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</table>
5. How many times per week do you exercise?
   _______ times/week
For how long each session?
   _______ minutes/session

6. What type of exercises do you do?
APPENDIX D

Informed Consent Forms

D.1 In Vivo Studies

SUBJECT AUTHORIZATION FOR INVESTIGATIONAL STUDIES
Lower-Leg Response to Impact Loads in 1G and Micro-G
(Part 1: In vivo Experiments)

1. Statement of Research

You are invited to participate in the research study as described below. It is a principle of medical practice that a subject who is to participate in the research investigation of a new medical treatment, device or procedure must give his or her informed consent to such participation. This consent must be based on an understanding of the nature and risks of the treatment, device or procedure. This document provides information important to this understanding. If you have any questions, please ask.

2. Information on the Research

The purpose of this study is to measure the forces acting on your calcaneus (heel bone) under different gravitational conditions (1G, zero-G, and simulated zero-G). Three kinds of measurements are needed (one of which is invasive). The first measurement involves performing vertical jumps on a force plate. The jumps will be "jumping jacks" or simple vertical jumps up in the air from both feet. The jumps will be performed intermittently with time to rest in between. A total of about 50 jumps will be performed. The force plate is similar in concept to a bathroom scale, except that weight is measured more rapidly and precisely. In addition to this, electrical activity of selected muscles of your lower leg will be recorded via sensors placed on the surface of the skin.

The invasive measurement involves two pins that will be surgically placed into your heel bone and relates to compression directly in the bone. These pins will be implanted by James Sferra, M.D., a Cleveland Clinic physician who has specialized in ankle and foot surgery. A portable fluoroscope unit will be used to ensure the proper position of the pins prior to insertion. Fluoroscopy is a radiographic method similar to xray which allows images to be acquired almost instantaneously. A fluoroscopic image of the calcaneus will be taken and the skin on the heel will be marked with a pen to indicate the intended position of the pins. The implantation requires local anesthesia only. After the anesthesia is applied, a small incision will be made in the skin, and a threaded intracortical pin (2 mm in diameter) will be inserted (and later removed) using a surgical drill (no predrilling is required). The length of time the pins will be in place will be about 2 hours. Previous
studies have been performed in our laboratory in which pins were similarly placed in subjects' knees or shoulders. In these studies, subjects reported no discomfort during insertion of the pins. In the present study, the area crossed by the pins does not involve any joints, muscle, or major nerves or blood vessels. We do not anticipate there being any permanent changes to bone or skin as a result of these procedures.

The gravitational conditions will be as follows. Zero-G conditions will be obtained in a special KC-135 aircraft used by NASA to study the effects of weightlessness. It flies from Johnson Space Center and the flight path consists of a number of parabolic trajectories (following an oscillating, broadly u-shaped path). Each flight lasts about 90 minutes. **Only force plate data will be collected during these flights.** Simulated zero-G will be studied by suspending subjects from latex cords, such that each limb segment is supported independently. When subjects are positioned in the simulator, they lie horizontally and face the ceiling. (Jumping is achieved by pushing against a wall.) The final gravitational condition (1G) simply involves subjects performing vertical jumps on a force plate that is mounted on the ground. Both types of force measurements (force plate and pin) will be recorded in the simulator and in 1G.

Participation in the study will require four visits -- one (one-day) visit to NASA Lewis Research Center in Cleveland (for pre-flight physicals), two (overnight) visits to Johnson Space Center in Houston (one for physiological training that is required before a person can fly on the KC-135, and one for the actual KC-135 flight), and finally one four-hour visit to the CCF for the 1G and simulated zero-G studies.

3. **Risks and Discomforts**

All of the types of measurements and gravity conditions have been studied before with minimal risk to subjects. The zero-G experiments are likely to cause some degree of motion sickness, but NASA does have some guidelines for minimizing this discomfort. These guidelines involve pre-flight dietary recommendations, opportunity to take anti-motion sickness medication, instruction on in-flight behavior to minimize motion sickness, and airsickness bags. The dietary and in-flight precautions will be explained in more detail during the training session.

The *in vivo* experiments do carry a slight risk of temporary localized pain while the pins are inserted or removed, and/or infection in the heel region, but strict sterile procedures will be followed to minimize this risk. To be implanted in the bone, the pins pass only through skin and fatty tissue, not through any muscle, joint, or major artery, vein, or nerve. You will be exposed to a very small amount of radiation from the fluoroscope. The amount of radiation will be well below that of a standard xray.

In the case of measuring muscle activity, the skin sites need to be lightly abraded prior to electrode application. These sites may experience some slight discomfort when alcohol swabs are used to clean the area.

There is a small risk associated with the two round-trip commercial flights from Cleveland to Houston for the training and KC-135 flights. This risk is the same as that associated
with any domestic flight on a commercial airline. There is also a small risk of the KC-135 aircraft crashing. However, none of these data-gathering flights has ever resulted in a crash, and the aircraft is flown by trained personnel.

During the KC-135 phase of the study, once a subject is aboard the plane, he or she must remain aboard while the flight is in progress. However, he/she may discontinue participation in data collection at any time during the flight.

4. Benefits

Individuals in this study will derive no direct benefits. Participation in the study will, however, provide investigators with valuable information concerning bone and muscle mechanics in altered gravity conditions as well as in normal conditions.

5. Alternative Procedures of Treatment

The study is for data-collection purposes only. Therefore, the alternative is to not participate in this study.

6. Confidentiality

Confidentiality of your records will be maintained; however, the Food and Drug Administration and the sponsoring organizations (NASA) may inspect the research records if needed. It is expected that data will be published in the scientific literature and/or presented at national research meetings. However, only group or aggregate information will be published.

7. Research-Related Injuries

If physical injury occurs due to involvement in research, medical treatment is available, but you or your insurance company must pay the cost of treatment. Compensation for lost wages and/or direct or indirect losses is not available. The Cleveland Clinic shall not provide compensation for medical expenses or any other compensation for research-related injuries. Further information about research-related injuries is available from the office of the Institutional Review Board (216/444-2924).

8. Questions about Research

If you have any questions about the research or develop a research-related problem, you should contact Brian Davis, Ph.D., at 216/444-1055. During non-business hours you should contact Dr. Davis (216/646-9787). Should Dr. Davis not be available after hours, Dr. Jim Sferra will be on-call and you may contact him by calling The Cleveland Clinic (216/444-2200) and asking for Dr. Sferra to be paged. If neither Dr. Davis nor Dr. Sferra is available, you should call 216/444-2200 and ask for the orthopaedic surgery resident on call. If you have any questions about your rights as a research subject, you should contact the Institutional Review Board (216/444-2924).

9. Voluntary Participation

Your participation in this study is voluntary. Your refusal to participate will not prejudice your future treatment or benefits here at The Cleveland Clinic. You are free to
discontinue participation in the study at any time without fear of penalty or loss of medical care. However, during the KC-135 phase of the study, once a subject is aboard the plane, he or she must remain aboard while the flight is in progress. However, he/she may discontinue participation in data collection at any time during the flight. If significant new findings develop during the course of the study which may affect your willingness to participate, you will be informed.

10. Costs

You are not responsible for any additional cost as a result of the research. You will receive $300 in remuneration upon completing this study, unless you are listed as a Co-Investigator on the grant which is funding this project.

Date Subject Signature

Date Witness Signature
D.2 Simulated Zero Gravity Studies

SUBJECT AUTHORIZATION FOR INVESTIGATIONAL STUDIES
Lower-Leg Response to Impact Loads in 1G and Micro-G
(Part 2: Simulated Zero-G Experiments)

1. Statement of Research
You are invited to participate in the research study as described below. It is a principle of medical practice that a subject who is to participate in the research investigation of a new medical treatment, device or procedure must give his or her informed consent to such participation. This consent must be based on an understanding of the nature and risks of the treatment, device or procedure. This document provides information important to this understanding. If you have any questions, please ask.

2. Information on the Research
The purpose of this study is to investigate jumping exercises in simulated zero-G and in normal gravity. The simulated zero-G condition will be studied by suspending subjects from latex cords, such that each limb segment is supported independently. When subjects are positioned in the simulator, they lie in a horizontal position and face the ceiling. (Jumping is achieved by pushing against a wall.)

Three kinds of non-invasive measurements are needed. The first involves performing vertical jumps on a force plate. The jumps will be "jumping jacks" or simple vertical jumps up in the air from both feet. The jumps will be performed intermittently with time to rest in between. A total of about 50 jumps will be performed in each gravitational condition. The force plate is similar in concept to a bathroom scale, except that weight is measured more rapidly and precisely. In addition to this, electrical activity of selected muscles of your lower leg will be recorded via sensors placed on the surface of the skin. Thirdly, the impacts of the jumps will be measured using a tiny (weighing less than 5 grams) accelerometer taped to the skin on the front of the shin (the antero-medial tibial surface).

The second and final gravitational condition (1G) simply involves subjects performing vertical jumps on a force plate that is mounted on the ground. The same three kinds of measurements will be recorded.

This study will involve a single visit to The Cleveland Clinic Foundation, lasting about 2 hours.

3. Risks and Discomforts
The measurements in this study involve standard biomechanical procedures. The time spent in the simulator will be approximately 80 minutes, and during this time some minor discomfort may be experienced where the straps of the harness press against the skin. If this occurs, every attempt will be made to place extra padding in the area of discomfort.

In the case of applying electrodes to the skin, some initial preparation involves lightly abrading the area to decrease skin resistance. This may be mildly irritating when an alcohol swab is used to clean the area.
4. Benefits

Individuals in this study will derive no direct benefits. Participation in the study will, however, provide investigators with valuable information concerning bone and muscle mechanics in altered gravity conditions as well as in normal conditions.

5. Alternative Procedures of Treatment

The study is for data-collection purposes only. Therefore, the alternative is to not participate in the study.

6. Confidentiality

Confidentiality of your records will be maintained; however, the Food and Drug Administration and the sponsoring organizations (NASA) may inspect the research records if needed. It is expected that data will be published in the scientific literature and/or presented at national research meetings. However, only group or aggregate information will be published.

7. Research-Related Injuries

If physical injury occurs due to involvement in research, medical treatment is available, but your insurance company must pay the cost of treatment. Compensation for lost wages and/or direct or indirect losses are not available. If you have any questions about your rights as a research subject, you should contact the Institutional Review Board (216/444-2924).

8. Questions about Research

If you have any questions about the research or develop a research-related problem, you should contact Brian Davis, Ph.D., at (216) 444-1055. During non-business hours you should contact Dr. Davis at (216) 646-9787. Should Dr. Davis not be available after hours, Dr. Jim Sferra will be on-call and you may contact him by calling The Cleveland Clinic (216/444-2200) and asking for Dr. Sferra to be paged. If neither Dr. Davis nor Dr. Sferra is available, you should call 216/444-2200 and ask for the orthopaedic surgery resident on call. If you have any questions about your rights as a research subject, you should contact the Institutional Review Board (216/444-2924).

9. Voluntary Participation

Your participation in this study is voluntary. Your refusal to participate will not prejudice your future treatment or benefits here at The Cleveland Clinic. You are free to discontinue participation in the study at any time without fear of penalty or loss of medical care. If significant new findings develop during the course of the study which may affect your willingness to participate, you will be informed.
10. **Costs**

You are not responsible for any additional cost as a result of the research.

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APPENDIX E

Data Collection Forms

E.1 Anthropometric Measurements

Measurements for Zero Gravity Simulator

Dimensions for ZGS

Floor to Calf COG: ___________ (cm) Calf to Thigh COG ___________ (cm)
Thigh COG to Waist: ___________ (cm) Waist to Chest: ___________ (cm)

Body Height: ___________ (cm)
15% Height: ___________ (cm)

Body Weight: ___________ (lbs) ___________ (kg)

Distance from Hip to COG
37.2% Thigh Length ___________ (m)
Midthigh Circumference: ___________ (m)

Calf Length: ___________ (m)

Distance from Knee to COG
37.1% Calf Length ___________ (m)

Calf Circumference: ___________ (m)
Knee Diameter: ___________ (m)

Foot Length: ___________ (m)
Malleolus Height: ___________ (m)
Malleolus Width: ___________ (m)
Foot Breadth: ___________ (m)

ZGS Measurements

Segment Mass
Lower Leg ___________ (kg)
Thigh ___________ (kg)

Distance of Support to Joint
Thigh (hip joint) ___________ (m)
Calf (knee joint) ___________ (m)
Lower Leg ___________ (m)
Thigh ___________ (m)
Force in Latex Cord

Force in Cord = \frac{\text{mass of segment} \times \text{distance hip to COG}}{\text{distance hip to support}}

ZGS Measurements

Angle between gravity replacement system rope and body

Front Angle \quad ________ (deg) \quad Back Angle \quad ________ (deg)

Length of springs

Front Spring \quad \begin{array}{c} 45\% \text{ BW} \\ 60\% \text{ BW} \\ 75\% \text{ BW} \\ 90\% \text{ BW} \\ 100\% \text{ BW} \end{array} (\text{cm}) \quad \begin{array}{c} 45\% \text{ BW} \\ 60\% \text{ BW} \\ 75\% \text{ BW} \\ 90\% \text{ BW} \\ 100\% \text{ BW} \end{array} (\text{cm})

E.2 Data Collection Forms

Data Collection Form

Subject Number: \quad ______________

Complete Informed Consent \quad ______________

Complete Health History \quad ______________

LabVIEW Data Directory: c:\labview\zgs\subject12

Motion Data Directory:

ZGS Data \quad /home6/aging/susan/cr12

Motion Data Directory:

1G Data \quad /home6/aging/susan/cr12a

Force Plate Data Directory: c:\dandrea\cr12

Channels

0 \quad \text{Calcaneal Strain}

1 \quad \text{Gastrocnemius EMG}

2 \quad \text{Vastus Lateralis EMG}

3 \quad \text{Calcaneal Acceleration}

4 \quad \text{Tibial Acceleration}

5 \quad \text{Force}

6 \quad \text{Synchronization Pulse}

7 \quad \text{Miscellaneous}
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### Motion File Convention

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APPENDIX F

Castigliano’s Theorem

Castigliano’s Theorem was named after the Italian engineer Alberto Castigliano (1847-1884) and can be used to determine the contribution to the strain due to the acceleration of the pins at impact. Since the strain in the bone is relatively small, any vibration in the pins may introduce large errors in the strain readings due to the displacement of each pin. By using Castigliano's Theorem, the deflection of the pins at the point of the strain transducer can be determined by computing the partial derivative \( \frac{\partial U}{\partial P} \) of the strain energy, \( U \), with respect to the load, \( P \) (Beer and Johnston, 1981).

In the current investigation, the load experienced by the pins is a product of the mass of the pin, transducer, base and target (Table F.1) and the acceleration measured by the calcaneal accelerometer.

<table>
<thead>
<tr>
<th>Component</th>
<th>Mass (g)</th>
<th>Mass (kg)</th>
<th>Length (cm)</th>
</tr>
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<tbody>
<tr>
<td>Pin</td>
<td>2.31</td>
<td>0.00231</td>
<td>9.525</td>
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<tr>
<td>Target</td>
<td>0.85</td>
<td>0.00085</td>
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<tr>
<td>Base and Sensor</td>
<td>1</td>
<td>0.001</td>
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**Table F.1:** Characteristics of the Pin, Target, Base and Sensor

The calcaneal pin was assumed to be a cylindrical cantilever beam (Figure F.1). Measurements were recorded from one subject to determine the length and mass of the pin as well as the position of the target, base and sensor on the pin (Table F.2), a critical
factor in order to determine the strain at the point where the base and target contact the pin. For the subject in question, long pins were placed in the calcaneus at the time of insertion. After the 1G trials, the pins were cut to a shorter length and the strain recorded during the simulated microgravity jumping experiments. The overall distance from the subject's skin to the end of the pin was recorded in both cases but for consistency, only the latter trials were analyzed using Castigliano's Theorem.

\[
\begin{align*}
\text{P}_{\text{TARGET}} & \quad \text{L} \\
\text{L}_0 \quad \text{x} \quad \\
\text{L} \quad \text{A} \quad \\
\text{W}_{\text{PIN}}
\end{align*}
\]

Where:
- \( L_0 \) is the distance from the calcaneus to the target (m)
- \( L \) is the length of the pin protruding from the calcaneus (m)
- \( P_{\text{TARGET}} \) is the mass of the target (N)
- \( W_{\text{PIN}} \) is the mass of the pin protruding from the calcaneus (N)

Figure F.1: Calcaneal Pin Represented as a Cantilever Beam for Castigliano's Theorem

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<th>Long</th>
<th>Short</th>
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<td>Distance from calcaneus to target, ( L_0 ) (m)</td>
<td>0.01499</td>
<td>0.01499</td>
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<tr>
<td>Length protruding from calcaneus, ( L ) (m)</td>
<td>0.0708</td>
<td>0.0264</td>
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<tr>
<td>Mass of length protruding from bone (g)</td>
<td>1.72</td>
<td>0.642</td>
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<tr>
<td>Weight of pin protruding from bone, ( W_{\text{PIN}} ) (N)</td>
<td>0.0169</td>
<td>0.0630</td>
</tr>
</tbody>
</table>

Table F.2: Measured Parameters Regarding Length and Mass of Calcaneal Pin
In the case of a beam, the strain energy is equal to:

\[ U = \int_0^L \frac{M^2}{2EI} \, dx \]  

(F.1)

where \( U \) is the strain energy
\( M \) is the moment
\( E \) is the modulus of elasticity
\( I \) is the mass moment of inertia
\( L \) is the length of the rod

Taking the partial derivative of Equation F.1, the deflection of the pin, \( x \), at the point of the strain transducer, \( j \), is determined by:

\[ x_j = \frac{\partial U}{\partial p_j} = \int_0^L \frac{M \partial M}{EI \partial p} \, dx \]  

(F.2)

The properties used in Equations F.1 and F.2 can be found in Table F.3.

Additional information required was the weight of the pin. The weight was determined by using the measured mass and the maximum acceleration of the calcaneus as measured by the calcaneal accelerometer. Using Equation F.2, the deflection of both pins was determined individually as well as the overall displacement measured by the strain transducer mounted on the pins.

<table>
<thead>
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<th>Properties of Stainless Steel Bone Pin</th>
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<tr>
<td>Modulus of Elasticity (Pa)</td>
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<tr>
<td>Mass Moment of Inertia (m^4)</td>
</tr>
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<td>( I = 0.25 \pi r^4 )</td>
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<tr>
<td>Radius (m)</td>
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<td>Mass per unit length (kg/m)</td>
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**Table F.3:** Properties of Stainless Steel Pin
APPENDIX G

Model Verification by Mathematical Derivation

In order to assure that the ADAMS program was predicting valid results for the rheological model of the bone, a mathematical derivation was performed and the results compared to those from ADAMS. The model contains a Hookean spring (B) and a Kelvin unit (A) composed of a spring and viscous damper (Figure G.1). For any force input to the model, the stress in both the A and B units is equal. This however, does not hold true for the strain due to the action of the damper. The stress and strain in the system for a given input is:

\[
\sigma_A = \sigma_B \quad \text{G.1}
\]

\[
\varepsilon_{\text{total}} = \varepsilon_A + \varepsilon_B \quad \text{G.2}
\]

![Figure G.1: Model of Bone. A - Kelvin Unit, B - Hookean Spring](image)

Using Equation G.1 and G.2, the stress strain relationships for this system can be developed. Converting this relationship into the \(s\) domain with Laplace transforms, the strain for a given input can be derived (G.3):
\[ \sigma_A = k_1 \epsilon + \eta \epsilon \]
\[ \sigma_A = k_1 \epsilon + \eta \epsilon \]
\[ \sigma_A = (k_1 + \eta \epsilon) \]
\[ \sigma_B = \frac{\sigma}{k_2} \]
\[ \epsilon_T = \epsilon_A + \epsilon_B \]

\[ \epsilon_T(s) = \left[ \frac{1}{k_1 + \eta s} + \frac{1}{k_2} \right] \sigma(s) \]

\[ \epsilon(s) = \left[ \frac{k_1 + k_2 + \eta s}{k_2 k_1 + k_2 \eta s} \right] \sigma(s) \]

G.3

For a given activity, the external stress can be measured. Therefore, for the bone system, the stress is set as a sinusoidal function and its Laplace transform determined (G.3).

\[ \sigma(t) = M \sin(\omega t) \]
\[ \sigma(s) = \frac{M \omega}{s^2 + \omega^2} \]

G.4

Substituting G.4 into the stress-strain relationship in G.3, strain in the s domain is derived (G.5).

\[ \epsilon(s) = \left[ \frac{k_1 + k_2 + \eta s}{k_2 k_1 + k_2 \eta s} \right] \frac{M \omega}{s^2 + \omega^2} \]

G.5

This equation can be solved using partial fractions which converts equation G.5 into three separate parts (G.6).

\[ \epsilon(s) = \frac{A}{s + \frac{k_1}{\omega}} + \frac{B s}{s^2 + \omega^2} + \frac{C \omega}{s^2 + \omega^2} \]

G.6
Equating Equations G.5 and G.6, the constants A, B and C can be solved and using the inverse Laplace transform, the strain can be expressed in the time domain for the stipulated input function (G.7).

\[ \varepsilon(t) = A e^{-k_j t} + B \cos(\omega t) + C \sin(\omega t) \]

\[ A = \frac{M \omega \eta}{k_1 + \omega^2 \eta^2} \]

\[ B = -\frac{M \omega \eta}{k_1 + \omega^2 \eta^2} \]

\[ C = \frac{M[k_1^2 + k_1 k_2 + \eta^2 \omega^2]}{k_2[k_1^2 + k_1 k_2 + \eta^2 \omega^2]} \]

Equation G.7 can be used to solve for bone strain given a sinusoidal input. The stiffness constants must be determined and the magnitude of the peak force acting on the bone. This type of model was considered to be ideal for the current investigation in which the relationship between external force and internal bone strain was desired. This model was implemented on the ADAMS software and the simulation results in ADAMS were compared to the calculated results. The magnitude of the strain predicted by either method was identical.
BIBLIOGRAPHY


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