Effect of isokinetic resistance training on ulnar stiffness in young, college-aged women

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(ABSTRACT)

Bone mineral content (BMC) and bone mineral density (BMD), measured by dual x-ray absorptiometry are used clinically to diagnose osteoporosis and estimate risk for fragility fractures. Bone mineral explains up to 70% of bone strength; however, it does not take into account bone geometry. Mechanical Response Tissue Analysis is a method of non-invasively measuring the bending stiffness (EI) of bone which is determined by the product of Young's modulus of elasticity (E) and the areal cross sectional moment of inertia (I). The aim of the current study was to determine if high intensity strength training will increase ulnar bending stiffness in young women. Forty-nine women aged 19.9 ± 1.7 yrs, trained their non-dominant arm either concentrically or eccentrically in the Isokinetic modality on the Biodex® system III 3d/wk for 32 wks. The dominant arm served as the control limb (untrained). Analysis of all subjects regardless of training mode demonstrated a significant increase in ulnar EI (22% ↑, P=0.01) with no significant difference in the untrained arm. When EI results were assessed by training mode, subjects who trained eccentrically showed a significant increase for ulnar EI in the trained limb (40% ↑, P=0.01) with no significant effect on the untrained limb while concentric training demonstrated no significant gain in either the trained or untrained arm. There was no effect of time x mode of training interaction for either the trained or untrained limb. Bone mineral density and bone mineral content of the ulna increased significantly in the trained arm in both concentric and eccentric training modes ($P<0.05$). These findings suggest support for the hypothesis that a critical threshold of mechanical bending loads may be necessary to effect an adaptation in bone strength and thus, eccentric training may be a novel approach to increase ulnar EI in young women.

Keywords: Mechanical response tissue analysis, bending stiffness, ulna, bone mineral content, bone mineral density

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Chapter 1

Introduction

Identifying Individuals at Risk of Osteoporosis and Fragility Fractures

The ability of bone to resist fracture is of increasing importance in our society. The risk of bone fracture increases proportionately with age and ~1.3 million fractures occur due to age every year (Bouxsein and Marcus, 1994). This; however, will likely worsen over time because the number of individuals over the age of 65 is expected to double by the year 2020, suggesting two or three times the number of age related fractures (Bouxsein and Marcus, 1994). A major limitation in the ability to assess the resistance of bone from fracture is the methods currently available to measure bone properties. The accepted clinical procedure to assess fracture risks is dual x-ray absorptiometry (DXA). This device estimates bone mineral content (BMC g/cm) and bone mineral density (BMD g/cm^2). The measurement of bone mass has developed into the clinical standard for diagnosing osteoporosis and predicting fracture risk (Watts, 2002). However, while BMD is an integral component of bone strength, this value is inadequate to completely predict fracture risk because bone strength is related to factors beyond mineralization. DXA can accurately measure the mass of cortical and trabecular bone; however, a major limitation in DXA technology is that a discernment between these two bone types cannot be determined (Sievänen et al, 1994). Cortical bone serves as a covering for trabecular bone in the axial

skeleton and is known to be stronger than trabecular bone. Therefore, a determination between these two types of bone is crucial before bone strength values can be accurately assessed. This factor is just one which represents a major limitation in predicting bone strength.

Total bone mass is associated with ~80% of bone strength (Sievänen et al, 1994). This leads to the theory that other factors, independent of BMC and BMD, affect bone strength. Martin stated bone strength consist of two fundamental properties: (i) shape and size of bone, and (ii) mechanical properties of bone (Martin, 1991). Mechanical properties of bone contain several parameters: composition, organization, collagen distribution, and damage already present to the bone and surrounding tissues (Beivier et al, 1989). Watts stated that fracture risk may fluctuate given age, gender, and race although BMD may be identical in each case (Watts, 2002). Also, Steele and colleagues stated that the mineral values, determined by photon absorptiometry, although indispensable are not ample for providing stiffness in bone (Steele et al, 1988). Roberts and colleagues address the issue that many studies find a poor correlation between mineralization of bone and bending stiffness or that other factors are more accurate predictors of bone failure properties (Roberts et al, 1996). Martin outlined that photon absorptiometry has too great a variation to be considered accurate and reliable to predict fracture risk. Photon absorptiometry has explained 48-88% of the variability in structural strength measurements (Martin, 1991). This;

however, is too variable to be considered a gold standard for assessing fracture risk. With this in mind, there must logically be other factors that influence bone strength in addition to the mineralization of bone.

New method of assessing bone mechanical properties

Since bone material can be determined by several methods all of which assess a material property, a device is needed to accurately measure the geometric properties of bone (i.e., how bone mineral is distributed about the central axis). Also, this measurement should be performed at low stress to the subject and possibly coupled with bone mineral values to accurately predict fracture risks. An experimental measure of bending stiffness (EI), which is composed to two properties: E = Young's Modulus of Elasticity (a material measure) and I = area moment of inertia (a geometric measure), is becoming more common to determine fracture risk. Bending stiffness, measured in Nm^2, is calculated as $K_b = EI/L^3$ where K_b is the transverse bending stiffness and L is the length of the bone being assessed. Elasticity is the resistance of bone to bending and will increase as BMC is distributed further from the central axis. Bending stiffness has been shown to be a good predictor in determining bone strength (Myburgh et al, 1993). Roberts and colleagues determined that stiffness is a more robust predictor of fracture risk than BMD alone; this is due to the fact that EI assesses both a geometric and material property while bone mineral values assess solely a material property (Roberts et al, 1996).

Mechanical Response Tissue Analysis (MRTA) is a device which assesses the bending stiffness (EI) of bone *in vivo*. Whealan states that the mechanical properties of a structure rely on the rigidity of that structure, which includes both material properties as well as the geometric properties (Whealan et al, 2000). He further goes on to discuss how any process employed to determine fracture risk must be able to measure both material and geometric properties of that structure (Whealan et al, 2000). This statement is the basis for the composite beam theory upon which MRTA is based. Martin stated that the mechanical properties of bone are subject to the identical laws upon which load bearing structures are governed (Martin, 1991). However, there is a major factor which bone possesses that makes it inherently unique in that it can adapt its architecture to loads imposed (Martin, 1991). Until recently, the only direct measure of bending stiffness could be performed *ex vivo* by three point bending in which the bone is broken and EI is measured directly. MRTA does not give an actual, representative three dimensional view of bone but does measure directly the bending stiffness of the bone in question, assessing both geometric and material properties of bone. Mechanical Response Tissue Analysis allows the operator to formulate a hypothesis abut how bone material is distributed around a central axis. Petersen and Orne found a strong relationship between fracture of bone and EI ($r = 0.96$) as well as with BMC ($r = 0.90$) (Orne et al, 1977). Myburgh and colleagues demonstrated that EI has a positive

correlation with BMC (r = .81) (Myburgh et al, 1993). However, this is likely not higher due to bone stiffness depending not only on mineralization but also on the geometry and elasticity of bone (Myburgh et al, 1993). Developed by Dr. Charles Steele, Stanford, in conjunction with NASA, the MRTA has several properties that make it unique. Foremost, the device can assess an experiment measure of the resistance of bone to bending (EI), *in vivo*. Also, this device can be operated with little to no discomfort to the subject. Finally, the MRTA provides for relative ease of operation for the technician. The MRTA along with the 6-parmeter prediction model has been validated by Roberts and coworkers (Roberts et al, 1996).

The MRTA is fitted with a probe, positioned directly at the midpoint of the bone in question which vibrates at a frequency of 60 to 1600 Hz, eliciting minimal discomfort. The feedback provided from the probe/shaker is then relayed to a signal analyzer, then to a computer program where a determined prediction algorithm calculates EI for that bone. MRTA also gives a sufficiency (S) score, which is important in determining bone strength properties. Sufficiency is simply load capability of the bone (P_{cr}) adjusted to body weight (BW) and size (P_{cr} /BW) (Steele et al, 1988). Sufficiency makes EI a relative value based on the unique characteristics of each individual subject. Simply stated, S determines the number of body weights that a bone can withstand in axial loading (Steele et al, 1988). With these variables in mind, MRTA is

able to measure different characteristics of bone and; therefore, can be used in adjunct with BMC scores to give a better representation of true bone strength (Steele et al, 1988).

There are several models to consider when bending stiffness is assessed. The main difficulty of assessing EI via vibration technique is the layer of soft tissue, or skin (Steele et al, 1988). Therefore, the models are based on parameters that "factor in" certain variables which are deemed crucial in determining bone stiffness. Two models have been validated for the ulna and are based on parameters that affect EI measures. The first parameter developed for the human ulna is the 7-parameter model. This model factors in the following variables: 1) mass of skin, 2) stiffness of skin, 3) damping of skin, 4) mass of bone, 5) stiffness of bone, 6) damping of bone, and 7) parallel damping of soft tissue (Steele et al, 1988). This model assesses EI relatively well for the ulna but encountered less than desired results for the tibia (Roberts et al, 1996). This is due to both intrinsic (biological factors related to organization of bone, distributions and restraints associated with surrounding connective and soft tissue) as well as extrinsic factors (positioning of limb, placement of probe, probe contact with bone, etc.) that are not taken into account fully by the 7 parameter model. Therefore, a second model was developed which was a refinement of the 7-parameter model. The new 6-parameter model included: 1) mass of skin, 2) stiffness of skin, 3) damping of skin, 4) mass of bone, 5) stiffness of

bone, 6) damping of bone (Roberts et al, 1996). With this model, parallel damping of soft tissue is included in damping of skin parameter (Roberts et al, 1996). Roberts and colleagues demonstrated a higher correlation when comparing EI determined by MRTA and EI determined by three-point bending using the 6-parameter vs. the 7-parameter demonstrating correlations of 0.947and 0.645, respectively (Roberts et al, 1996). This 6-parameter model; however, assumes the bone is fixed, or clamped at each end and thus, not allowing for free vibration which does not account for how a long bone behaves *in vivo*. For instance, bone is not fixed at each end completely and does, in fact, vibrate freely with only physiological restraints (i.e., collagen, ligaments, and the surrounding musculature) at the proximal and distal ends. Therefore, a third model was theorized, the 9-parameter model, which allowed for free vibration at either the proximal or distal end of bone. This 9-parameter model includes the variables measured in the 6-parameter model of skin and bone but also factors in bone mass, damping, and stiffness on either the distal or proximal end of bone. A final 12-parameter model was developed which included mass, damping, and stiffness scores at the distal and proximal ends of bone as well as mass, damping, and stiffness of bone and skin to equal 12 parameters measured. This final model mimics closely how bone behaves *in vivo*.

To aid in fracture risk prevention, intervention must be applied at the time of or before peak bone mass accumulation. Several studies have

attempted to evaluate the benefits of exercise to increase BMC and BMD in females under the age of 30. Friedlander and coworkers stated that the goal of initiating exercise habits in youth was to prevent fracture risk after menopause due to increased bone mass attained while young (Freidlander et al, 1995). Black and colleagues demonstrated that higher BMD decreased the risk of fracture late in life (Black et al, 1991). By adding more mineral content to bone early on, the loss of bone after menopause will not be as severe due to earlier addition of bone material. This is not to say less bone will be lost as aging occurs but that more bone material attained while young will lead to a greater concentration of mineralization in older bones compared to those who did not participate in exercise early in life. Friedlander and coworkers showed that after two years of aerobic strength training, there was a 0.53 % increase in BMD while women who only participated in stretching exercise suffered a 1.85 % decrease in BMD (Freidlander et al, 1995). This evidence demonstrates that bone loading is one of the keys to increasing mineral content of bone in youth and, thus, possibly preventing fracture risks as older adults. Some studies have found a non-significant increase in BMC and BMD throughout training; however, a major increase in bone stiffness occurs (Turner, 2003). This is due to the distribution of BMC throughout bone, remodeling mainly on the medial and periosteal surfaces where bone stress is the greatest (Turner and Robling, 2003). To this extent, small increases in BMC can, in fact, produce major gains in total bone strength.

Bone Adaptive Responses to Variations in Mechanical Loading

Wolff's law states that the internal and external architecture of bone tissues have the unique ability to augment when loads are imposed and stress occurs (Wolff, 1892). Dynamic loading of bone has been shown to produce the greatest increases in new bone growth as opposed to static loading (Turner and Robling, 2003). Turner described that dynamic movements create hydrostatic pressure, which initiates the movement of extracellular fluid within bone (Turner and Robling, 2003). As this fluid moves from an area of high concentration to an area of low concentration, pressure is exerted on osteocytes (bone cells), osteoblasts (bone forming cells), and bone lining cells (Turner and Robling, 2003). This pressure change creates shear stress on osteoblasts (Turner and Robling, 2003). Accordingly, shear stress is a major motivating factor for bone cells to begin new bone formation by stimulating osteoblasts to lay down "new bone" (Turner and Robling, 2003). Martin defines that bone responds to mechanical loading in two distinct ways (Martin, 1991). First, the shape and size of bone are impacted on the external areas of bone, namely the periosteum and the endosteal ends of bone, while material changes take place in the internal area of bone, especially the trabecular and Haversian bone areas (Martin, 1991). Without the proper stressor, bone will eventually resorb and lose its mineral content.

Frost coined the term "mechanostat theory" which refers to osteocytes being able to detect strains produced by the muscle to either

turn on or off the remodeling of bone (Chilibeck et al, 1996). The purpose of this system is to maintain the homeostatic nature of both bone deformation and stiffness while at maximal exertional effort of about 2000 microstrains (Chilibeck et al, 1996). This system works on negative feedback; therefore, if the musculoskeletal system is not stressed and bone has no stimulus to remodel, it will indeed resorb. Ehrlich and Lanyon stated that prolonged immobilization of limbs will result in decreased bone mass and also bone mineral alterations (Ehrlich and Lanyon, 2002). According to the mechanostat theory, this change would be due to less musculoskeletal stimulation to bone and therefore osteocytes "turning off" bone remodeling (Chilibeck et al, 1996). Ehrlich proposed that bone adapts based on stimuli provided by normal everyday activity (Ehrlich and Lanyon, 2002). He suggests that since bone fractures are caused by very intense impacts sustained by bone, then bone can not adapt to very high loads due to bone failure and therefore must remodel to a far less magnitude of stimulation (Ehrlich and Lanyon, 2002).

In theory, this system is off when normal strain occurs on bone tissue but is activated in any situation external to the normal range (Ehrlich and Lanyon, 2002). However, Ehrlich dictates that this theory may be limited in the ability to accurately define bone remodeling (Ehrlich and Lanyon, 2002). It is well known that bone adapts to dynamic loads imposed upon it; however, bone does not respond at all to

static loads and may even resorb if static loads are imposed on bone over time (Ehrlich and Lanyon, 2002). Clearly there must be other factors that help define bone remodeling other than a simple on/off switch. Schaffler and colleagues demonstrated that bone under normal physiological conditions (0 and 1500 με) does experience bone fatigue but maintained its integrity after a number of loading cycles were applied (Schaffler et al, 1990). This leads to the belief that even though bone is fatigued, which is defined as the loss of elasticity, failure may not inevitably occur. Only after very intense strains, above that of normal physiological levels, will bone tend to fail.

Resistance Training and Bone Strength

Strength training is effective for increasing bone strength in numerous ways. Chilibeck and colleagues stated strength training is beneficial for bone strength in three ways: 1) it creates dynamic versus static loads on bone, 2) strength training increases the strain on bone, and 3) the strain is not placed on one area of the bone (Turner and Robling, 2003). Bouxsein and Marcus describe that strength training may benefit women in several ways (Bouxsein and Marcus, 1994). First, resistance training may increase the amount of bone material early in life therefore making loss of bone mass later in life less stressful (Bouxsein and Marcus, 1994). Also, as women age, estrogen levels naturally decrease which is associated with quick bone loss; therefore, resistance training may lessen this bone loss (Bouxsein and Marcus, 1994). Lastly,

resistance training in older individuals may slow age related loss of bone mass, decrease the risk of falling, and attenuate the harshness of the fall (Bouxsein and Marcus, 1994). They came to the conclusion that exercise may provide increases in the strength of the skeleton but resistance training must be maintained throughout life in order to maintain these gains (Bouxsein and Marcus, 1994).

Many studies have assessed the effects of training on bone properties. However, there are very few studies that have measured the effects of training on variables other than BMC and BMD. It has been shown that muscle mass is positively correlated with bone strength. Doyle and colleagues found a high correlation with bone mass and muscle mass in male runners (Doyle et al, 1979). Also, Beivier and coworkers found a positive correlation between back strength and lumbar spine BMD (Beivier et al, 1989). It may be; therefore, correct to presume that training of the arm will increase muscle mass of the arm and henceforth ulnar bone mass of the arm. If this hypothesis is correct, EI will also increase due to the results shown above.

Statement of the Problem

The purpose of the current study is first, to assess if high intensity training will increase ulnar EI, BMC and BMD, and second, to determine the extent to which training will increase peak torque of the trained arm and if the change in arm strength, BMC, and BMD can accurately predict a change in EI.

Significance of the Study

The ability to predict fracture risk in young women is crucial to bone health of the elderly. Bouxsein and colleagues stated there are about 1.3 million age-related fractures annually (Bouxsein and Marcus, 1994). With this in mind, it is imperative that the bone health of our youth is improved in order to possibly deter bone fracture in later life. Currently the gold standard of fracture risk prediction is absorptiometry. However, this method is flawed by the fact that it assesses only a material property of bone by estimating the relative amounts of bone content (Sievänen et al, 1994). However, in order to obtain a clear representation of bone strength, the geometry of bone must also be assessed. Sievänen and associates stated that bone mass accounts for roughly 80% of bone strength (Sievänen et al, 1994). The geometry of bone relates to how bone content is distributed around a central axis. Roberts and colleagues reported that bone stiffness, a geometric property, may be a more robust predictor of bone strength than BMD alone (Roberts et al, 1996). This is of significant clinical importance due to the MRTA not only able to assess EI, but also being relatively inexpensive, non-invasive, and eliciting little to no discomfort to the subject.

With this in mind, the current study was aimed to determine if high-intensity strength training would increase ulnar EI, BMD, and BMC. In addition, this study also attempted to determine if the change in peak torque could be employed to predict the change in ulnar EI, BMD, or

BMC. The findings of the study may have substantial benefits to women of all ages but particularly younger women.

Research Hypothesis

H_o (1): High intensity isokinetic resistance training will increase ulnar bending stiffness in young adult women to a degree that is related to the adaptive response of the limb to strength training.

H_o (2): High intensity resistance training will increase ulnar BMD and BMC to a lesser extent than ulnar EI.

H_o (3): Peak torque of the upper arm will be the most robust predictor of ulnar EI.

Assumptions

1) Subjects trained with maximum effort at each testing session.

2) Subjects were truthful in answering pre-training questionnaires.

3) The Biodex System III, MRTA, and DXA were accurate and precise in their measurements.

4) Subjects did not train outside of the protocol set by the T.i.B.I.A.L researchers.

5) Subjects were untrained with little to no prior training experience.

Delimitations

1) Participants were women between the ages of 18-26 who attended Virginia Tech or resided in the surrounding community.

2) Subjects were randomly assigned into either a concentric or eccentric training group.

3) Subjects were acclimated to the training protocol prior to commencement of exercise.

4) Participants possessed no diseases or consumed medications that are known to alter bone metabolism.

Limitations

1) All subjects did not attain 100% compliance.

2) All participants did not exert maximal effort at each testing session.

3) Adherence to protocol may not have been followed per outside activity.

4) Male subjects were not taken into consideration.

5) Results of the study are not indicative to the entire population of 18-25 year old females.

6) Subjects did not train in our laboratory during fall, winter, or spring breaks.

Definition of Terms

1) **Area moment of inertia-** denoted as (I), areal moment of inertia is defined as the allocation of bone material about its bending axis, which increases as bone content is spread farther from the central axis (Myburgh et al, 1993).
2) **Bending stiffness (EI)-** The product of Young's Modulus of Elasticity and the areal moment of inertia, measured in Nm^2, which demonstrates the resistance of a bone to bending (Myburgh et al, 1993).
3) **Bone modeling-** Normal process of laying down bone. This process does not require osteoblasts and osteoclasts working in unison. This is an independent process in which stresses imposed upon bone is not the leading factor in commencement of this process.
4) **Bone properties-** bone has two fundamental properties that influence

the strength of bone i) shape and size of bone, and (ii) mechanical properties of bone (Martin, 1991).

5) **Bone remodeling-** The process of adding bone to existing bone due to stimuli produced from loading of that bone.
6) **Cortical bone-** A very hard outer covering of every bone. This bone type is very resistant to bending stresses. Cortical bone gives strength to the midpoint of bone where bending is not a desired trait.
7) **Distal end of bone-** The end of the bone farthest from the point of insertion, i.e. near the carpals.
8) **Distal 1/3 site-** 1/3 the distance from the distal end of the bone relative to the entire bone length
9) **Dual X-ray Absorptiometry (DXA)-** Device which can estimate the total mass of both trabecular and cortical bone; however, DXA cannot distinguish between the two (Sievänen et al, 1994). This device can also estimate the relative amount of bone mineral content and bone mineral density
10) **Dynamic loading of bone-** Considered to be extremely osteogenic, dynamic loading consists of cyclic forces imposed on the bone.
11) **Geometric properties of bone-** how bone mineral is distributed about a central axis.
12) **Hydrostatic pressure-** Pressure created by the movement of fluids. In bone, osteocytes and osteoblasts are stimulated by the movement of fluid from an area of high concentration to an area of low

concentration caused by hydrostatic action (Turner and Robling, 2003).

13) **Mechanical properties of bone-** composition, organization, collagen distribution, and damage present to the bone and surrounding tissues (Martin, 1991).

14) **Mechanical Response to Tissue Analysis (MRTA)-** Device which noninvasively assesses bending stiffness.

15) **Mechanostat Theory-** A feedback system that regulates the relative amounts of bone material. Simply put, bone undergoing strains greater than 1500 microstrains will remodel adding bone material. However, bone in disuse, achieving strains less than 100 microstrains will eventually resorb leading to a loss of bone material (Turner, 1999).

16) **Nine-parameter model-** Evaluates the mass, stiffness, and damping of both skin and bone; however, also allows for free vibration at either the proximal or distal end of bone.

17) **Osteoblasts-** Specialized bone forming cells that initiate bone material deposition when stimulated (Aarden et al, 1994).

18) **Osteocytes-** Osteoblasts which have been trapped in bone matrix during remodeling of bone. There may exist as much as ten times as many osteocytes as osteoblasts in bone tissue (Aarden et al, 1994).

19) **Probe-** Sensor that is placed against a subjects skin allowing a measurement to be attained.

20) **Proximal end of bone-** Part of bone closest to the point of insertion. For example, the proximal end of the ulna attaches to the humerus.

21) **Seven-parameter model-** Evaluates the mass, stiffness, and damping of skin and bone as well as the parallel damping of skin.

22) **Shaker-** Apparatus that provides the vibration of the probe.

23) **Shear stress-** Stress imposed by forces pulling against each other. This type of stress can cause deformations due to the two forces acting against a plane.

24) **Signal analyzer-** Device on the MRTA which relays input from the shaker to a computer algorithm.

25) **Six-parameter model-** Evaluates the mass, stiffness, and damping of both skin and bone with parallel damping of skin included in damping of skin.

26) **Sufficiency (S)-** The load capability of the bone (P_{cr}) adjusted to body weight (BW) and size (P_{cr} /BW) (Steele et al, 1988).

27) **Trabecular bone-** The internal matrix, or scaffolding of bone. This bone has elastic properties and not as dense as cortical bone. Trabecular bone comprises mostly the bone tissue in the axial skeleton including the bones of the skull, ribs and spine.

28) **Twelve-parameter model-** Evaluates the mass, stiffness, and damping of skin and bone but allows for free vibration at both the proximal and distal ends of bone, mimicking bone *in vivo*.

29) **Wolff's law-** States that tissue, such as bone and muscle, will

compensate for the loads imposed upon them and remodel in response to the stresses imposed (Wolff, 1892).

30) **Young's Modulus of Elasticity-** The force necessary to elongate material. Elasticity is the ability of an object to resist a distorting object and return to its original shape and size when the stress is removed.

Summary

The resistance of bone to bending is comprised of more than simply the mineralization of bone. Sievänen and colleagues have documented that total bone mass has been shown to explain about 80% of bone strength (Sievänen et al, 1994). This leads to the theory that other variables influence bone strength in conjunction with mineralization. The manner in which the bone material is distributed may be more important than mineralization alone in predicting fracture risk. Bending stiffness has been shown by Roberts and colleagues to be a more accurate predictor of fracture risk than mineralization values alone (Roberts et al, 1996). Therefore, the MRTA was developed to determine how mineral content is arranged around the central axis, thus assessing a geometric property. Therefore, the aim of this research was to determine if high intensity isokinetic resistance training will increase EI in the ulna.

Chapter II

Review of Literature

Introduction

The current study was performed to determine if high-intensity isokinetic strength training will increase ulnar bending stiffness in young women. As a secondary purpose, to investigate if the change in upper arm peak torque can predict the change in ulnar bending stiffness (EI), bone mineral content (BMC), and bone mineral density (BMD). The principle function of this chapter is to provide the reader with a) risk of fragility fractures among women, b) the properties of bone which determine the strength of bone and the response of bone to loading c) current methods of assessing bone fracture risk d) a review on how the Mechanical Response Tissue Analysis (MRTA) assesses bending stiffness (EI) and sufficiency (S) and the rationale on why it is a useful tool to employ clinically for fracture risk prediction.

Risk of fragility fractures among women

Fragility fractures due to osteoporosis, or low bone mass, are a major health concern affecting our nation. As a point of reference, Cummings and colleagues state that one in every six white females will likely suffer a hip fracture in their lifetime (Cummings et al, 1995). Currently, Frost has proposed that there are biological or genetic determinants which could cause or aid in inducing osteoporosis (Frost, 2001). In this view, Frost emphasizes that osteopenia and osteoporosis

may be due to genetic factors but may be lessened with proper bone loading through exercise (Frost, 2001). Frost states there are four fundamental units associated with total bone strength in relation to fracture risk: 1) the material property of bone, 2) relative amount and type of bone present, since trabecular bone is stiffer than cortical bone, 3) the shape and size of bone, a geometric factor, and 4) damage already present in bone or microdamage previously induced (Frost, 2001). If these factors could be altered for the better through loading, bone may be less likely to develop into an osteopenic or osteoporotic state, even if genetic factors are involved. In fact, Friedlander and colleagues state the purpose of increasing activity at a young age is to deter fracture occurrence in the older population through a middle aged increase in bone mass (Friedlander et al, 1995). This group goes on to state that most research is focused on the older adult population; however, some researchers believe the premenopausal group, those aged 20-40 show the most promise in reducing the occurrence of osteoporosis in later life (Friedlander et al, 1995).

Bone properties, BMC and BMD, are an important aspect of determining the integrity of bone. A diminished bone density increases the chance of fracture with the estimated cost of fractures at $20 billion annually (Kelley et al, 2001). Faulkner identified that in many studies BMD values have been associated with fracture risk, in fact, a minute decrease in bone density of about 10-15% can virtually double the risk

for fracture (Faulkner, 2000). Cummings and coworkers identified several distinguishing traits that predispose women to suffering a fragility fracture. These investigators, in analyzing their data, found that smoking, decreased body weight below that of normal for age and height, high caffeine consumption, and physical inactivity were among the major risk factors for fracture in women (Cummings et al, 1995).

There are many studies published which have shown that bone mineral values are positively associated with fracture risk prediction (Hui et al, 1989: Cummings et al, 1993: Melton et al, 1993: Marshall et al, 1996). Melton and colleagues discovered that bone loss at the femoral neck is basically linear over the lifetime (Melton et al, 1993). Baseline and sixteen year follow up BMD at the femoral neck correlates to 0.83 (Melton et al, 2003). However, BMD at the femoral neck for fracture prediction past ten years is not established (Black et al, 2000). With this in mind, it is unclear if BMD is an acceptable determinant of fracture risk in younger individuals since there has been no evidence of the successfulness of this variable long term. This is of important clinical significance since a major goal is to decrease fracture occurrence in later life through bone improvements while young.

Cummings and colleagues conducted a prospective study from 1986 to 1988 to assess the incidence of hip fracture in white women (Cummings et al, 1995). During this study, they recruited 9516 white women over the age of 65 with no incidence of previous hip fracture or

hip replacement (Cummings et al, 1995). They also excluded black women because of their relatively low rate of hip fracture occurrence (Cummings et al, 1995). There was no intervention but the subjects were followed for about 4 years after initially agreeing to participate in the study (Cummings et al, 1995). Of the 9516 study participants, motor vehicle accidents caused 192 fractures of the hip, while 526 participants died, and 92 subjects dropped out of the study (Cummings et al, 1995). This group recognized 16 independent risk indicators for fracture risk besides bone density, which is displayed in Table 1 below (Cummings et al, 1995).

Table 1: Risk factors for fracture in Caucasian women and how these factors affect the chance for fracture.

Risk Factor	**Risk of Fracture above normal**
Age (per 5 years)	*1.5*
History of maternal hip fracture vs. none	*2.0*
Increase in weight since age 25 (per 20%)	*0.6*
Height at age 25 (per 6 cm)	*1.2*
Self-rated health (per 1 point decrease)	*1.7*
Previous hyperthyroidism vs. none	*1.8*
Current use of long-acting benzodiazepines	*1.6*
Current use of anticonvulsant drugs	*2.8*
Current caffeine intake (per 190 mg/day)	*1.3*
Walking for exercise	*0.7*
On feet ≤ 4 hr/day	*1.7*
Inability to rise from a chair	*2.1*
Lowest quartile for distant depth perception	*1.5*
Low-frequency contrast sensitivity	*1.2*
Resting pulse rate > 80 bt/min	*1.8*
Any fracture since age 50	---
Calcaneal bone density (per 1 SD decrease)	---

From this table, it is clear that there may be many risk factors for developing a hip fracture in white women. Each of these factors, with the exception of walking for exercise and height after age 25 demonstrates at least a 1 fold greater chance of suffering a hip fracture even without bone mineral factored into the equation. In addition, this group identified that the more weight gained past the age of 25, the less risk there is for fracture compared to those who lost weight which doubled fracture risk (Cummings et al, 1995). There are also four distinct physical traits which increased fracture risk: failure to get out of a chair without the use of the arms, an increased resting pulse, inferior visual perception of depth, and decreased low-contrast sensitivity (Cummings et al, 1995). Also, this group found that a decreased calcaneal bone density was associated with an increased risk of hip fracture (Cummings et al, 1995). As a drawback to the study, Cummings and colleagues pointed out several limitations which include no men in which to compare fracture risk, no younger women to determine if these same risk factors apply to the younger population, and no other ethnic groups in which to compare these results (Cummings et al, 1995). However, this study does indicate which risk factors, other than low bone mineral density, may be associated with fracture risk in older, post-menopausal women (Cummings et al, 1995).

Cummings and colleagues performed a study similar to the above in 1993 in which they assessed BMD at different skeletal sites to determine fracture risk (Cummings et al, 1993). From 1986 to 1988, 9704 Caucasian women were assessed for BMD and followed for a course of two years spanning to 1990 in which there were 8134 remaining subjects (Cummings et al, 1993). They measured BMD and BMC at the mid and distal forearm and calcaneus as well as the proximal femur and lumbar spine (Cummings et al, 1993). This group found that bone density measured at the hip was strongly associated with an increased risk of hip fracture, demonstrating a decrease of only 1 standard deviation relating to a 2 to 6 fold increase in fracture risk (Cummings et al, 1993). They also determined that BMC was less of an indicator than BMD to assess fracture risk of the hip (Cummings et al, 1993). Cummings and colleagues determined that for every increase in age by 10 years, there is a subsequent increase in fracture risk at the hip from 2 to 9 fold (Cummings et al, 1993). This study indicates that measurement of BMD is a good predictor of hip fracture in Caucasian women.

In conclusion, it is evident that the risk of fracture is associated with lifestyle factors as well as the mineralization of bone itself. However, it is also suggested that young women may have a higher bone mineralization than older, post-menopausal women. With this in mind, if the younger population is able increase the mineral content of bone while

in youth, this may effectively decrease the chance of fracture later in life through a decreased consequence of bone loss after menopause.

Properties affecting bone strength

Martin stated that bone strength is determined by two fundamental properties (i) mechanical properties of the composite material and (ii) size and shape of bone (Martin, 1991). The mechanical properties of bone include bone mineralization, organization of the material, relative location of the collagen, as well as the damage already present in bone (Steele et al, 1988). With this in mind, BMC and BMD values give no insight into the size and shape of bone, which is referred to as bone geometry. The geometry of bone indicates how bone material is distributed throughout the length of the bone. As bone is distributed along the periosteal ends, where stresses are the greatest, bone becomes stronger. In fact, very small increases in mineral content along the periosteal ends may result in great increases in the bending stiffness of that bone.

Bone integrity is difficult to quantify due to the rapidly dynamic activity within the tissue. To determine resistance of bone to bending, the bending stiffness (EI) of bone, an equation has been developed which incorporates the elastic modulus (E) as well as the areal moment of inertia (I). These parameters take into account both the material properties of bone as well as the geometric properties of bone. The elastic modulus, a material property, is simply the resistance of a bone to

bending. As (E) increases, bone is less likely to fracture due to greater stiffness within the composite material. The areal moment of inertia (I), a geometric property, describes how the bone is distributed about a central axis. With any hollow material, as (I) increases, the center of mass is distributed farther from the central axis allowing the material to be more resistance to breakage. The stress/strain curve (Figure 1), demonstrates how bone behaves when loads are applied.

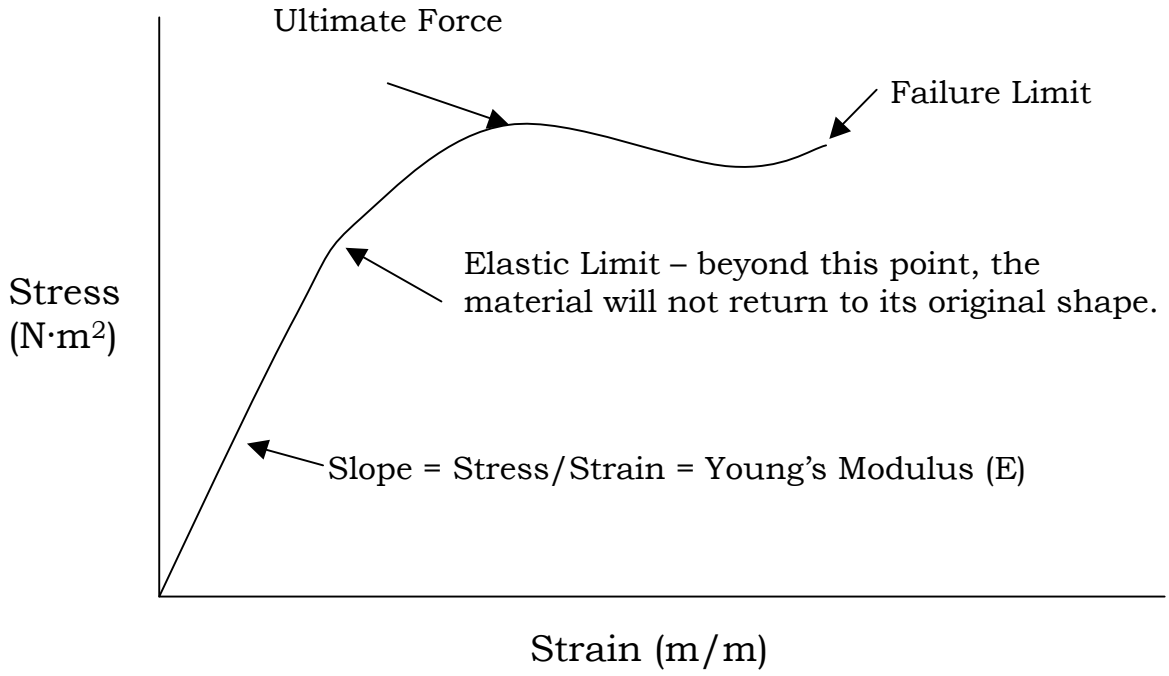

Figure 1: Stress/Strain curve representing the elastic, plastic, and failure regions of bone.

This diagram indicates that bone has an elastic region such that as stress is applied, bone will adapt and return to its normal origin. However, if bone is stressed beyond the elastic region, known as the plastic region, permanent damage will occur. Once the plastic region has

been reached, bone can no longer return to its normal shape and size due to microfractures which have occurred. If the stress is too great, beyond the limits of the plastic region, fracture of the bone will inevitably result due to failure of the material.

Bone behaves based on the stresses that are incurred. For example, in shear, tension, or compressional stress, bone behaves as $L_f = A\sigma_f$ where Lf is the load failure, σ_f is defined as the strength of the material, and A is the area in cross-section (Martin, 1991). For this equation, it is clear that as the area or the material strength increase so has to the failure moment and vice versa. Martin also defines $M_f = (I/c)\ \sigma_{bf}$ to describe the bending of bone. In this equation, M_f is the moment of failure, σ_{bf} is defined as the bending failure stress, (I) is the moment of inertia in cross-section, and c is the distance from the outer fibers (Martin, 1991). Here it is clear that the failure moment is greatly increased as areal moment of inertia and the size of bone are increased as the failure stress for bending, σ_{bf}, is a constant. These two equations help to explain what is necessary to increase both the load failure and the moment of failure in bone. In both instances, bone integrity is greatly enhanced by increasing the size of bone thus greatly reducing the risk of bone failure. It is improbable to determine the geometry of bone via DXA variables alone thus the MRTA is a unique tool which can be employed to determine, according to Martin, what determines bone failure moments.

Bone remodeling in response to the loading of bone

Wolff's law states that bone remodels as the result of loads being applied to bone, in which the bone will adapt to those loads placed upon it (Wolf, 1892). Bone adapts to those loads by increasing the mineral content as loads are applied and resorbing bone content as loads are removed over time (Snow-Harter et al, 1992). As bone is stressed under normal physiological conditions not great enough to invoke fracture, bone will remodel to accommodate the loads imposed upon it. Frost gives the representation of bone remodeling involving the Activation-Resorption-Formation theory (Frost, 1998). He states that basic multicellular units (BMU's) form osteoclasts, which eventually become osteoblasts, to fill tiny holes in bone with new lamellar bone (Frost, 1998). These holes within bone are termed the "remodeling space" and can inhabit up to 25% of a bone's area; however, usually only inhabit around 4% (Frost, 1998). These holes in bone are small and relatively insignificant unless bone has resorbed to a state were the number or size of the holes are too large for osteoblasts activity to adequately replace lost bone (Frost, 1998). Under this example, Frost illustrates that when BMU's are less active than the number of remodeling spaces, termed the disuse-mode model, losses may occur in bone and lead to an osteopenic state (Frost, 1998). This entire remodeling period usually takes around three months in duration to complete (Frost, 1998). With this in mind, it is not difficult to understand why or how bone will resorb if adequate loads are not placed upon it to initiate BMU activity. However, if BMU

activity is constant with the resorption state, termed the conservation-mode, then bone loss will be halted (Frost, 1998). As a theory on how bone remodels when loads are imposed, Frost coined the mechanostat theory of remodeling. Frost denotes bone remodeling is either turned on or off based on the stress that is applied to bone (Frost, 1998). The mechanostat theory of bone remodeling is based on use-disuse which is described by Sugiyama and colleagues. During restricted or no use, bone will resorb and the mass will decrease, which is known as the disuse mode (Sugiyama et al, 2002). In the adaptive phase, stresses are applied to bone great enough to hold bone mass loss/gain at homeostasis but less than that required for the remodeling threshold to be reached (Sugiyama et al, 2002). However, when stresses are applied to bone which exceed the remodeling threshold, bone mass will tend to increase, which is termed the overloading phase (Sugiyama et al, 2002). The mechanostat theory runs in parallel with the BMU theory by Frost in that as bone is stressed, remodeling is turned "on" and new bone is formed. However, when bone is not stressed at least to the point of conservation mode, disuse occurs and bone material is resorbed, rendering bones weaker and less dense. Frost states that stresses applied to bone beyond the remodeling threshold point of around 50-100 microstrains will adequately initiate the conservation process (Frost, 1998).

As a preventative measure, exercise has been shown to maximize bone development during early life in preparation for bone loss as aging occurs (Kelley et al, 2001). Heinrich and colleagues demonstrated bone mass, both in the axial and peripheral skeleton, were greater in younger women that participated in resistance training verses aerobic training (Snow-Harter et al, 1992). Resistance training imposes dynamic loads on bone not seen during aerobic training. Also, bone tends to respond to dynamic loading exclusively. Burr states that dynamic stresses are essential to promote bone growth and also to avert inhibition of bone development of the periosteal surface (Burr et al, 2002). Wolff's law indicates that bone will remodel with respect to loads placed upon it (Wolff, 1892). Accordingly, bone is stressed to a greater extent through resistance training as compared to aerobic activities. Turner and Robling proposed the following model, Figure 3, to show the rough cascade of events in which bone remodels (Turner and Robling, 2003).

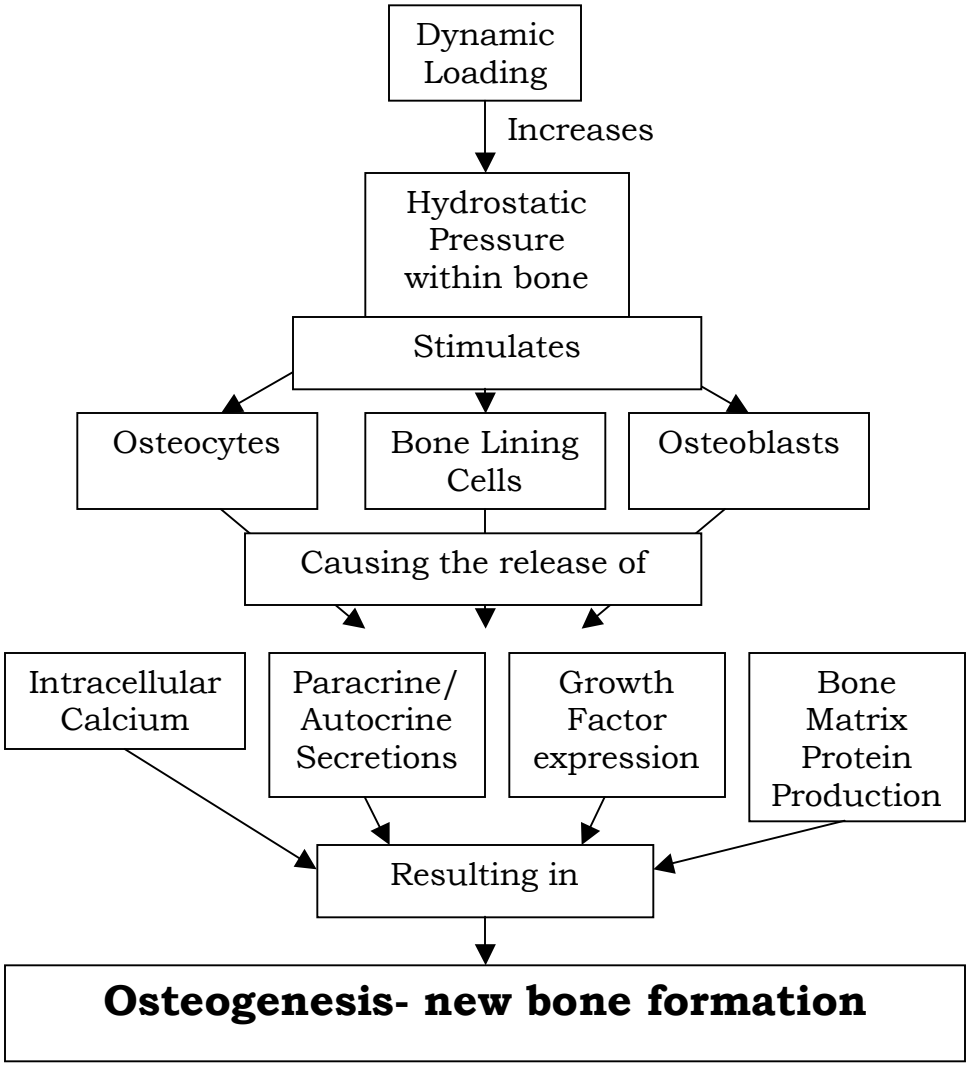

Figure 2. Diagrammatic cascade of factors associated with bone remodeling.

It is known that exercise is positively correlated with an increase in bone strength. Turner and Robling state that habitual exercise impacts the size, shape, and density of bone resulting in increases in the mechanical strength of bone (Turner and Robling, 2003). Resistance training, in particular, places increased stress on the bone which initiates a cascade of events, explained earlier by Turner, in which bone

material is added to the area of bone in which the stresses imposed are the greatest. Snow and colleagues found that collegiate gymnasts displayed an increased BMD of the hip and spine throughout an 8 month season but lost bone density as the 4 month off-season ensued (Snow et al, 2001).

Bone loading through resistance training should increase bone quality more so than other methods of mechanical stimulation. Friedlander and colleagues demonstrated this theory in a study performed in 1995 (Friedlander et al, 1995). This group studied 63 Caucasian and Asian American women between the ages of 20 and 35 over the course of two years (Friedlander et al, 1995). These women were divided into 4 groups: exercise + calcium, exercise + placebo, stretching + calcium, and stretching + placebo (Friedlander et al, 1995). The exercise group attended three sessions/week which incorporated weight training via dumbbells, barbells, and wrist or ankle weights (Friedlander et al, 1995). The subjects were asked to increase the weight used periodically throughout the two year program (Friedlander et al, 1995). The stretching group served as the control group which consisted of three classes/week for 30 minutes in duration (Friedlander et al, 1995). Based on the subject's daily calcium intake, calcium supplementation was regulated to provide the subjects with 1500 mg of daily calcium (Friedlander et al, 1995). At baseline, all groups were similar in their fitness parameters as well as bone mineral measurements (Friedlander et

al, 1995). However, at the completion of the study, both exercise groups demonstrated significantly higher BMD (0.53 ± 2.6%) than the stretching groups (Friedlander et al, 1995). In fact, the stretching groups actually lost BMD (1.85 ± 5.7%) throughout the study (Friedlander et al, 1995). This group demonstrated that resistance training is superior for increasing mineral status as compared to non-resistance exercise.

Snow-Harter and colleagues performed a study in which they assessed young women to determine the effect of resistance and endurance training on bone mineral status (Snow-Harter et al, 1992). This group recruited 52 women, of which 31 completed training, with a mean age of 20 yr and divided them into three groups: resistance training, endurance training, or control for a total of 8 months (Snow-Harter et al, 1992). Resistance training consisted of 14 exercises (total body) three times/week for 8-12 repetitions at each station (Snow-Harter et al, 1992). The participants were asked to increase the weight lifted as they grew stronger (Snow-Harter et al, 1992). The endurance group consisted of running at least three sessions/week and were instructed to add mileage each week as they grew more accustomed to training with an average of 2.3 miles/week at months 1 and 2 and 4.9 miles/week at months 7 and 8 (Snow-Harter et al, 1992). The control group maintained their normal activity level in which recreational activity participation was emphasized (Snow-Harter et al, 1992). The resistance trained and endurance trained increased BMD significantly from baseline with 1.2

±2.2% and 1.3 ± 1.6%; however, these exercise groups did not differ from each other statistically (Snow-Harter et al, 1992). No difference was found in the control group (Snow-Harter et al, 1992). From this study it is unclear whether resistance training is more a robust form of increasing bone mineralization than endurance training other than the resistance trained group had a greater increase in BMD.

Other researchers also have focused on determining the effects of resistance exercise on bone. Fujimura and colleagues found that a four month strength training regimen significantly increased serum osteocalcin, as well as serum bone-specific alkaline phosphatase levels in young (23-32 yr) Oriental males (Fujimura et al, 1997). These investigators also determined that urinary excretion of deoxypyridinoline simultaneously was suppressed (Fujimura et al, 1997). This suggests that bone formation occurred with suppression of bone resorption in the resistance trained group (Fujimura et al, 1997). Also, Nelson and colleagues found that high-intensity resistance training increased both femoral neck and lumbar spine BMD in women, early post-menopause (Nelson et al, 1994). Also, total body BMD was well preserved over the 1 yr training protocol in that study, but found to decrease in the control group (Nelson et al, 1994). Adami and colleagues found that following six months of resistance training employing exercises specifically designed to maximize the stress imposed on the wrist bones, no differences were detected in BMC between the group that trained and the one that did not

(Adami et al, 1999). However, cortical bone cross-sectional area and volumetric density increased at the ultradistal radius (Adami et al, 1999). These increases were found to be significant in comparison to the control group and thus, Adami and colleagues concluded that site-specific training producing only small deviations in bone mass may contribute important structural changes in the affected bone sites and in turn, increase the bending stiffness of bone (Adami et al, 1999).

It is evident that physical activity may increase bone mineralization and thus bone strength. However, what is unclear is what form of physical activity is superior for this purpose. Some researchers believe resistance training is key to incorporate into any program due to the effect this mode of training imposes on the mechanical loading of bone. It is true that resistance training may load bone to a greater degree than other, not-impact forms of exercise. With this in mind, resistance training should be far superior for an increased bone remodeling response thus enhancing bone strength compared to other forms of exercise.

Current methods of assessing bone structural properties

Clinically, dual x-ray absorptiometry (DXA) is the current gold standard for assessing bone material properties, i.e. BMD and BMC (Sievänen et al, 1994). Dual X-Ray Absorptiometry is also the current method of choice for diagnosing osteoporosis and osteopenia. Several studies have been conducted which demonstrate that a low BMD is

positively linked to fracture risk (Cummings et al, 1993; Ross et al, 1991; Melton, 1993). However, DXA is hindered in the ability to only to estimate the relative mineralization of bone without an actual, representative three dimensional view of bone that a computer tomography (CT) scan will provide. Also, the mineralization of bone is not the only factor affecting the integrity of bone. Sievänen and colleagues have explained that total bone mass accounts for roughly 80% of bone strength (Sievänen et al, 1994). Therefore, no technique which assesses bone mineral only can precisely predict long term fracture risk since about 20% of bone strength remains unexplained through bone mineral measurements.

There are essentially two methods that use ionizing radiation to assess bone material: computer tomography (CT) and photon absorptiometry. In the use of radiation to assess bone mineral properties, BMC and BMD can be estimated. This; however, is a somewhat misleading view of the bone integrity. Instruments which measure BMD, with the exclusion of quantitative computer tomography (qCT), determine the integral mass of bone; which is the combination of both cortical and trabecular bone (Faulkner, 2000). Steele and colleagues state that bone mineral measurements, while essential, are not adequate to accurately predict bone stiffness (Steele et al, 1988).

Dual X-Ray Absorptiometry has several characteristics which label it invaluable for a clinical setting. This device is able to measure a wide

variety of variables including whole body BMC and BMD, fat mass, and fat-free mass (Nagy and Clair, 2000). Photon absorptiometry is also relatively inexpensive when compared to CT or qCT scans which provide the same basic information.

Dual X-Ray absorptiometry was evaluated for precision in determining total body BMD by Nagy and colleagues in mice (Nagy and Clair, 2000). Twenty-five mice were scanned with the DXA three times, weighted, sacrificed and then dried to determine total bone ash composition (Nagy and Clair, 2000). The coefficient of variation (CV) for the total body BMD (g/cm^2) was determined to be 0.84% and 1.60% for total body bone mineral between all three measurements (Nagy and Clair, 2000). This group found that DXA determined bone mineral variables were highly correlated to those found by ash comparison with a correlation of .88 between total body bone mineral and chemical ash (Nagy and Clair, 2000). This correlation is improved to .98 if the single outlier was eliminated (Nagy and Clair, 2000). In conclusion, Nagy and colleagues determined that DXA is an accurate procedure to determine total body bone mineral due to the high correlation with mineral ash composition taken after sacrifice of the animals (Nagy and Clair, 2000).

Lochmuller and colleagues conducted a study to determine if other techniques such as peripheral quantitative computer tomography (pQCT) or quantitative ultrasound (QUS) could replace DXA in diagnosing osteoporosis (Lochmuller et al, 2003). This group measured bone

properties of 126 human cadavers at the hip, spine, and radius with DXA and the radius only with pQCT, also they determined bone properties at the calcaneus with QUS and the spine with QCT (Lochmuller et al, 2003). The bones were then exposed to stress to determine the failure points of the femur, thoracolumbar spine, and the radius (Lochmuller et al, 2003). At completion of the testing, DXA based measures explained 55% of the variability that occurred at the failure point of the femur while pQCT and QUS explained only 15-40% (Lochmuller et al, 2003). Ultrasound failed to give any additional insight as to the mechanical properties of any of the three sites measured (Lochmuller et al, 2003). They concluded that for a clinical setting to assess bone strength, femoral DXA measurements are indeed similar to pQCT or QUS (Lochmuller et al, 2003).

Lochmuller and colleagues performed a similar study to the one above in which they tested three techniques, DXA, QUS, and QCT to determine which would best predict the mechanical strength of the radius under different loading patterns (Lochmuller et al, 2002). This group measured DXA of the forearm, hip, spine, and total body; pQCT of the distal radius; QCT of the spine; and QUS of the calcaneus (Lochmuller et al, 2002). The left radius was subject to three-point bending as well as axial compression while the right radius underwent falling simulation (Lochmuller et al, 2002). Dual X-Ray Absorptiometry correlated with failure in three point bending, compression, and fall simulation with 0.89, 0.84, and 0.70, respectively (Lochmuller et al,

2002). They also found that correlations involving pQCT were not greater than those reached using DXA (Lochmuller et al, 2002). Lochmuller and colleagues concluded that site specific DXA is the best clinical indicator for diagnosing the loss of bone strength compared to pQCT, QCT, or QUS (Lochmuller et al, 2002).

Mechanical Response Tissue Analysis

The Mechanical Response Tissue Analysis (MRTA) is a relatively new technique employed to assess the bending stiffness (EI) of bone *in vivo*. This device was developed by Dr. Charles Steele in conjunction with NASA and assembled by Gait Scan (Gait Scan Inc., Ridgewood NJ). The MRTA measures two unique properties of bone; bending stiffness, which is the resistance of bone to bending, and sufficiency (S), which is the load carrying capacity of that bone (Steele et al, 1988). Bending stiffness is determined by the bone's response to a low frequency vibration (Steele et al, 1988). The MRTA is composed of three functional devices: the shaker, the probe, and the signal analyzer. Roberts outlined the procedure in which the probe is positioned at the midpoint of the ulna where the shaker emits a low frequency vibration, usually around 60-1600 Hz, propelling the probe against the subject's skin (Roberts et al, 1996). This real time data is streamed to a signal analyzer which models the data to the parameter model chosen by the operator. The shaker, probe, sensors, and platform from which the MRTA is supported are illustrated in Figure

4 taken from our laboratory (Laboratory for Health and Exercise Science, Virginia Tech).

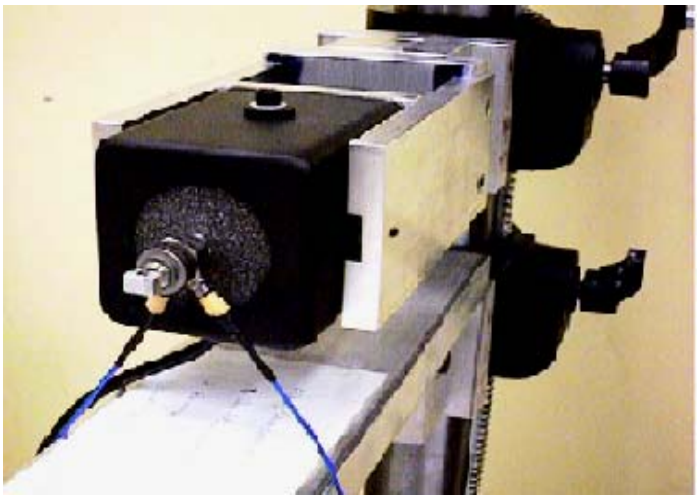

Figure 3. MRTA device: shaker, probe, sensor, and platform.

When assessing EI, Steele denotes that the bone behaves as a beam, the skin as a spring, and the surrounding musculature provides massive damping (Steele et al, 1988). The main difference in the MRTA versus other vibratory methods is the assessment of the data collected. Instead of using the response curve, the impedance curve, or the climax of any point on the curve, which is the resonant frequency, MRTA technology employs the entire response curve to quantitatively derive the mechanical properties of the bone (Steele et al, 1988). To illustrate the advantage of this method, Steele cites Doherty and colleagues in the comparison of osteoporotic bone with two normal controls (Steele et al, 1988). Doherty found that the variation in resonant frequency of the osteoporotic and normal bone was only 20%; however, the resonant frequency difference between these two bone states were 80% different by

way of bending stiffness (Doherty et al, 1974: Steele et al, 1988). In this example, the resonant frequency did not explain the difference between the two bone types to a large extent as did bending stiffness (Steel et al, 1988). Steele denotes that resonant frequency is determined by the ratio of stiffness to mass; however, in osteoporotic bone, both stiffness and mass decrease (Steele et al, 1988).

Bending stiffness is determined mathematically from a series of parameters. Steele gives the formula $k_b = [(48) \cdot (EI)]/L^3$ in order to determine the bending stiffness of the bone in question (Steele et al, 1988). In this equation, k_b is the lateral bending stiffness and is used to define a supported homogeneous beam with force directed at the midpoint (Steele et al, 1988). The lateral bending stiffness is used to determine EI due to the fact that the ulna behaves as a uniform beam in bending (Steele et al, 1988). The other factor associated with this equation is the length of the bone to be assessed (Steele et al, 1988). In this, it is clear that a slight mistake in ulna measurement, from the styloid process to the olecranon process, can result in major error in the actual EI measurement being that the value of the length is cubed. Since bone behaves as a uniform beam in bending, this method of determining EI is based on a beam composed of a uniform cross section to determine an averaged EI (Steele et al, 1988).

To date, two models have been validated for use by the MRTA. The first model proposed by Steele and colleagues in 1988, was the seven

parameter model (Steele et al, 1988). This model accounts for the mass, stiffness, and damping of the skin and bone but also includes parallel damping of soft tissue (Steele et al, 1988). Parallel damping is a separate entity due to the effect of soft tissue damping being the greatest difficulty to overcome in assessing bending stiffness (Steele et al, 1988). Accordingly, this hardship is what sets the MRTA apart from other methods to assess mechanical stiffness, in which the soft tissue component is taken into account to determine the bending stiffness (Roberts et al, 1996). However, this method elicited mixed results. The ulna performed beautifully but the tibia achieved less than desired results (Roberts et al, 1996). Therefore, Roberts and colleagues devised a refinement of the former 7-parameter model to the 6-parameter model (Roberts et al, 1996). In this model, parallel damping is included in the damping of skin variable (Roberts et al, 1996). The rationale behind this change is due to the skin being an incompressible soft tissue layer, which Roberts proposed could be incorporated into the damping of skin variable (Roberts et al, 1996). For validation, Roberts and colleagues used 12 Rhesus monkeys in which EI was determined by both MRTA and 3-point breakage (Roberts et al, 1996). Both the 7-parameter and 6-parameter models were incorporated to determine EI via MRTA and then compared to direct breakage using 3-point bending (Roberts et al, 1996). Roberts found EI by way of the 6-parameter model ($R^2 = 0.947$) to be far superior in predicting the fracture point versus the 7-parameter model

(R^2 = 0.645) when compared to *ex vivo* breakage determined by 3-point bending (Roberts et al, 1996). Both the 6- and 7-parameter models consequently performed well when measuring EI of the ulna (Roberts et al, 1996). This may be due to the fact that the ulna is intrinsically less likely to bend due to less length than the tibia (Roberts et al, 1996). The tibia is longer than the ulna and perhaps makes it naturally more flexible eliciting the poor results with the 7-parameter model (Roberts et al, 1996). However, there is a major limitation to the 6-parameter model. This model assumes that the bone is clamped at each end allowing for no free vibration (Roberts et al, 1996). However, this is not how bone behaves *in vivo*. Tendons, ligaments, and musculature that surround bone hold the bone in place while allowing for free vibration. With this in mind, two other models were conceived based on the 6-parameter model that do allow for free vibration at the ends of bone and still incorporate parallel damping in the damping of the skin parameter. The 9-parameter model assumes the same mass, stiffness, and damping of skin and bone but also accounts for the mass, stiffness, and damping of either the proximal or distal end of bone. Next, a 12-parameter model was developed that went one step further in assessing how bone behaves *in vivo*. The 12-parameter model allows for free vibration at both the proximal and distal ends of the bone. The major limitation of the 9- and 12-parameter models, thus far, are no direct validation as with the 7- and 6-parameter models that are currently accepted. However, to the

author's knowledge, there are current validation studies being performed which involve the 12-parameter model.

Sufficiency is defined as the amount of body weights the bone can carry without fracture occurrence (Steele et al, 1988). Steele outlined that the two most important properties the MRTA determines is the bending stiffness (EI) and the load carrying capacity of bone (S) calculated as P_{cr}/BW (Steele et al, 1988). Sufficiency is a dimensionless variable with enormous clinical importance since this measure allows us to determine how many body weights a bone can support in axial loading without the risk of fracture. With the load carrying capacity of a bone determined, an index may be produced that will lead to a measure of that bones integrity. Average S has been determined in canine ulna as 6.7, monkey ulna as 12, and human ulna as 7.1 (Steele et al, 1988). It is clear that S is similar in different species based on these values. Unlike EI, which may differ greatly according to the anatomical considerations within bone itself, S is a parameter which is based on the body weight of the subject (Steele et al, 1988). Steele and coworkers found S closely related to two variables, obesity and activity level (Steele et al, 1988). Apparently those individuals who were obese or that maintained sedentary lifestyles posed a very low S score (Steele et al, 1988).

Roberts and colleagues outline that there have already been several studies performed to assess bone strength of long bone *in vivo*, in which vibration response has been the foremost focus (Roberts et al,

1996). Since bone mineral measurements lack the ability to determine the essential geometric properties of bone, the MRTA is employed in order to assess the geometric distribution of BMC as the result of an impedance curve, which can then be incorporated with bone mineral data to further enhance fracture prediction (Roberts et al, 1996).

Myburgh and colleagues performed a cross-sectional training study employing 51 healthy men aged 28-51 where activity level was the key focus on ulnar bending stiffness to illustrate the effectiveness of exercise on bone bending stiffness (Myburgh et al, 1993). Three groups were classified into highly active (H), moderately active (M), and sedentary (S), based on retrospective classification and physical activity history (Myburgh et al, 1993). Results show that biceps and grip strength values were very similar for both the M and S groups but significantly higher for the H group ($P < .0001$) (Myburgh et al, 1993). Also, BMD of the ulna was significantly higher for the H group ($P < .05$) compared to the M and S groups (Myburgh et al, 1993). Additionally, ulnar EI was significantly higher for the H group ($P < .01$) than the M or S groups which did not differ significantly (Myburgh et al, 1993). Myburgh and colleagues employed the 7-parameter model to assess EI in these subjects which was, as stated earlier, sufficient for use in assessing the ulna (Myburgh et al, 1993). The foremost findings of this study demonstrate that activity level is highly correlated with ulnar bending stiffness (Myburgh et al, 1993). Secondly, Myburgh was able to demonstrate that the MRTA is an

applicable tool to determine that a higher level of exercise may strengthen bone compared to moderate or no exercise at all (Myburgh et al, 1993).

Kiebzak and colleagues performed a study to show how ulnar EI decreases in osteoporotic Caucasian women (Kiebzak et al, 1999). In this study, DXA variables were used to determine normal bone, osteopenic bone (classified as a T score flanked by -1 and -2.5), and osteoporotic bone (classified as a T score below -2.5) (Kiebzak et al, 1999). One hundred and twenty-five caucasian women were assessed and placed into three age groups: less than 40 yr, between 41-60 yr, and those women at or over the age of 61 (Kiebzak et al, 1999). Results for this study are presented in Table 2 below (Kiebzak et al, 1999).

Table 2: Results from Kiebzak et al, in assessing ulnar EI and bone mineral values based on the classification of bone.

	Classification of Bone		
	Normal	**Osteopenic**	**Osteoporotic**
Ulnar Bending Stiffness	33.1 ± 6.9	30.1 ± 8.7	25.0 ± 4.9
BMD (1/3 site)	0.713 ± 0.074	0.659 ± 0.093	0.570 ± 0.069
BMC (1/3 site)	1.41 ± 0.187	1.32 ± 0.144	1.11 ± 0.136

In addition to Table 2, Kiebzak also found a significant age related decrease in ulnar EI when women of the oldest age group were compared to the younger two age groups (Kiebzak et al, 1999). As indicated from Kiebzak et al, bone integrity decreases as the bone becomes less dense, or towards an osteoporotic state (Kiebzak et al, 1999). With this in mind, the MRTA is a tool capable of determining the relative fracture risk of

young bone in order to modify the present activity level to compensate for age related bone loss and bone fragility. Kiebzak's group concluded that osteoporotic bone possessed a significantly lower EI than that of osteopenic or normal bone, with no significant difference between these latter two bone types (Kiebzak et al, 1999). Therefore, if bone integrity can be maintained through physical activity, then loss of critical bone mass may be deterred and, in effect, lessen the risk of fragility fractures due to low bone strength.

Summary

In summary, osteoporosis is a serious concern for the women of our nation. This condition weakens bone to the point that fracture may occur even when only moderate trauma is experienced. However, bone can be strengthened by loading via physical activity, specifically, resistance training. In this case, bone mineral can be conserved throughout life due to a greater amount of bone mineralization that occurred as a result of training while accumulating bone mass at a young age.

The current method to diagnose osteoporosis is by way of DXA. This device estimates the relative amount of bone mineral at different sites as well as over the entire body. While this method has been validated to assess bone mineral properties, it fails to accurately predict long term fracture risk due to bone possessing strength properties beyond mineralization alone (Martin, 1991). Bone has two properties: (i)

material properties, which DXA can measure and (ii) the geometric properties, which DXA can not assess (Martin, 1991). The geometry of bone can be explained in how the bone mineral is distributed around a central axis. Since DXA can not give a three dimensional view of bone, an alternate device is necessary in order to assess that portion of bone integrity.

A relatively new, non-invasive device is being employed for fracture prediction risk. This device is termed the Mechanical Response Tissue Analysis (MRTA). The MRTA estimates the material properties of bone, as the modulus of elasticity (E), and the geometric properties of bone, as the areal moment of inertia (I). With these two properties of bone accounted for, the bending stiffness is estimated and thus should give an excellent index for bone fracture risk since both the material and geometric properties of bone are assessed. Possibly more importantly, this device give a measurement of sufficiency (S) which determines how many body weights a bone can carry without fracture occurrence. This variable has many clinical implications in that this one measure can tell the clinician how strong the bone may be and the likelihood of fracture if trauma is incurred.

Chapter III

Effect of isokinetic resistance training on ulnar stiffness in young, college-aged women

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Abstract

Bone mineral content (BMC) and bone mineral density (BMD), measured by dual x-ray absorptiometry are used clinically to diagnose osteoporosis and estimate risk for fragility fractures. Bone mineral explains up to 70% of bone strength; however, it does not take into account bone geometry. Mechanical Response Tissue Analysis is a method of non-invasively measuring the bending stiffness (EI) of bone which is determined by the product of Young's modulus of elasticity (E) and the areal cross sectional moment of inertia (I). The aim of the current study was to determine if high intensity strength training will increase ulnar bending stiffness in young women. Forty-nine women aged 19.9 ± 1.7 yrs, trained their non-dominant arm either concentrically or eccentrically in the Isokinetic modality on the Biodex® system III 3d/wk for 32 wks. The dominant arm served as the control limb (untrained). Analysis of all subjects regardless of training mode demonstrated a significant increase in ulnar EI (22% ↑, P=0.01) with no significant difference in the untrained arm. When EI results were assessed by training mode, subjects who trained eccentrically showed a significant increase for ulnar EI in the trained limb (40% ↑, P=0.01) with no significant effect on the untrained limb while concentric training demonstrated no significant gain in either the trained or untrained arm. There was no effect of time x mode of training interaction for either the trained or untrained limb. Bone mineral density and bone mineral content of the ulna increased significantly in the trained arm in both concentric and eccentric training modes (P<0.05). These findings suggest support for the hypothesis that a critical threshold of mechanical bending loads may be necessary to effect an adaptation in bone strength and thus, eccentric training may be a novel approach to increase ulnar EI in young women.

Keywords: Mechanical response tissue analysis, bending stiffness, ulna, bone mineral content, bone mineral density

INTRODUCTION

Bone fracture due to decreasing bone mass is a major health concern, especially among women. To assess fracture risk, the current gold standard is to use the normative standard for bone mineral content (BMC g/cm) and bone mineral density (BMD g/cm^2) by Dual energy X-Ray Absorptiometry (DXA) (1). However, bone mineralization alone is not sufficient to fully predict fracture risk (1). Martin outlined that bone has two inherent properties which influence strength: 1) the size and shape of bone and 2) the properties of bone material (2). The material properties can be estimated via DXA; however, bone geometry, or how the bone material is distributed around a central axis, cannot be accounted for using this technique. Cummings and colleagues have documented that roughly one in six white women will likely suffer a hip fracture in their lifetime (3). Several studies have already shown that bone mineral values strongly correlate with fracture prediction risk (4,5,6,7). However, as stated, bone mineral values are not adequate enough to explain the resistance of bone to fracture.

The resistance of a bone to fracture is directly related to the strength which can be measured as the product of the elastic modulus (E) and the areal moment of inertia (I) (2). Bone tends to fail or fracture when loads are applied which exceed the limits for the bone. A experimental device has been developed termed the Mechanical Response Tissue Analysis (MRTA), which estimates the mechanical strength as it is

affected by both the material and how it is distributed about the bone (1,8). Developed by Dr. Charles Steele, Ph.D. in conjunction with NASA, the MRTA has the means to estimate, in vivo, the properties from which bone strength are derived. This device employs low-frequency vibration (60-1600 Hz) elicited through a probe positioned perpendicular to the midpoint of the bone. The feedback is subjected to a 6-, 7-, 9-, or 12-parameter model to estimate the bending stiffness (EI) of a bone. There are two components which comprise EI: Young's modulus of elasticity (E), a material property; and the areal cross sectional moment of inertia (I), a geometric property. Roberts and colleagues validated the MRTA via measuring actual 3-point breakage values against MRTA derived values (1). This group demonstrated a correlation between EI_{MRTA} and $EI_{3\text{-point bending}}$ of 0.947 with the 6-parameter model and 0.645 using the 7-parameter model, both employing monkey tibia (1).

Exercise has been proposed as a method of increasing bone mass in individuals. Aloia and colleagues have demonstrated that bone mass is higher in male marathon runners than in males who are sedentary (9). Also, Dalen and colleagues have shown that BMC is higher in those who are physically active compared to inactive individuals (10). Myburgh and colleagues assessed bending stiffness of the ulna and BMC from 51 healthy male subjects (28-61 yr) and compared them across three groups: sedentary (n=13), moderately active (N=18), and highly active (N=20) individuals, according to self-reported activity levels (11). This was a

cross-sectional study design with no training protocol; this was done in order to determine the relationship between ulnar EI and activity level without age bias (11). This group found that the highly active group demonstrated statistically significant maximal bicep strength, grip strength, ulnar BMC, and ulnar EI compared to the sedentary and moderately active groups whom where similar with each other (11). Also, grip and bicep strength measurement had a significant positive correlation with activity level; however, activity level did not correlate with bone mineral values (11). This group did determine that muscular strength did correlate significantly with EI and BMC and that EI correlated with BMC and ulnar width (11). Interestingly, ulnar width and strength of the bicep independently predicted EI (11). Myburgh and colleagues concluded that ulnar EI can be influenced by activity levels adequate enough to increase arm strength (11).

Strength training, as a form of high-intensity exercise, should elicit a response more than adequate to stimulate bone growth and thus exert a positive effect on ulnar bending stiffness as well as BMC and BMD. Therefore, the primary purpose of this study was to determine if high-intensity muscular strength training increases ulnar bending stiffness in young, college-aged females. A secondary aim was to determine which training mode is superior for eliciting increases in bone stiffness.

METHODS

Subjects

Forty-nine subjects aged (Mean ± SD: 19.9 ± 1.7 yr) in the following study were assessed for height (163.3 ± 6.1 cm) and weight (59.7 ± 8.7 kg) at baseline. Measurements throughout the study were obtained on the dominant and non-dominant arm and therefore, to assess arm dominance, subjects were asked which hand they would use to throw a ball to mark the dominant arm. Subjects participating in the current study were recruited from a larger Trial in Bone Injury Abatement for Ladies (T.i.B.I.A.L) database of 125 subjects. However, of these 125 subjects who participated in the early phase of the training, only 80 completed the protocol and were measured for outcomes at the end of the study. Furthermore, only subjects with a root mean square (RMS) error of less than 10% for measurement of ulnar bending stiffness were accepted as subjects included in the final analysis. Of the 80 subjects who completed training, 49 meet the RMS < 10% inclusion criteria for final analysis and those who did not meet this criteria were found to not differ in body weight, height, age, or fat mass from those who did meet the RMS criteria. The data of each subject included in this final group was analyzed using the 12-parameter model to assess ulnar EI. This model accounts for the mass, stiffness, and damping of soft tissue and bone, as well as the mass, stiffness, and damping at the proximal and distal ends of bone which seems to most accurately mimic how bone behaves *in vivo*.

Subjects were evaluated for total ulna BMC (g/cm) and BMD (g/cm^2) at weeks 0 and 32 and bending stiffness ($N \cdot m^2$), total work of the arm (ft·lbs), and peak torque of the arm at weeks 0, 16, and 32 (ft·lbs). However, no subjects in the current study were assessed for these variables at week 16 which was a time point intended solely for the larger T.i.B.I.A.L study. Subjects were asked not to begin an exercise program or if exercising currently, to continue only light exercise throughout the duration of the 32 week study. All training and testing sessions occurred with the subject rested and relaxed.

Strength training

Only the non-dominant arm was trained throughout the duration of the study and thus the dominant arm served as the control. Biceps and triceps peak torque values were determined using the Biodex System III®. Subjects were randomly assigned to a concentric or eccentric high-intensity training group in order to maximize the training effects of the trained limb. Each subject performed a low intensity set of six repetitions to acclimate in response to high-load resistance training. Subjects training consisted of three sessions per week employing five sets of six repetions each, for a maximum effort with a latency period of 45 seconds between sets. Subjects began acclimating to training three weeks before baseline with one set of six repetions. The following week, they progressed to three sets of six repetions for three non-consecutive days. Finally, one week before training formally began; they completed

five sets of six repetitions in preparation for the rigorous training protocol. Alternate exercises were performed by all subjects while academic activities were halted during fall, winter, and spring break.

Strength testing

Subjects were tested at baseline (wk 0) and post-training (wk 32) for concentric and eccentric peak torque and total work. During testing, there were three separate strength measures obtained including maximum isometric strength (elbow at 90º), maximum flexion, and maximal extension for each arm. Testing consisted of one maximal concentric and one maximal eccentric set of six repetitions for each arm followed by one set of three maximal isometric repetitions for each arm.

Ulnar Bending stiffness

Ulnar bending stiffness (EI) was determined via MRTA. Nine measurements were taken per limb by the MRTA for each subject to ensure a quality measurement was obtained. Subjects reported to each session relaxed and non-stressed and were seated with their testing arm held at 90º. The subjects positioned their hand lightly around a secure bar to prevent movement while testing was performed. The elbow was allowed to rest on a pedestal with a soft padded surface. The measurement was taken at the midpoint of the ulna, which was defined as half the distance between the olecranon and styloid processes. The probe was position on the crest of the ulna perpendicular to the midpoint and the subjects remained still for a series of nine measurements with

15 seconds between measures. Each measurement took roughly 1 minute to perform for a total of 12 – 15 minutes of testing time per limb. The device was calibrated every three hours or whenever the machine was turned on. The shaker, which vibrates the probe, oscillates at a frequency of 60-1600 Hz. Feedback from the bone is then relayed to a signal analyzer and then to the computer where bending stiffness is calculated. Bending stiffness, or EI, is calculated by the equation $EI = k_b \times L^3/48$. To assure a good measurement was attained, a RMS of 10% was the limit for data inclusion. If any data obtained revealed a RMS greater than 10%, subsequent data for that subject pre and post were discarded. All data were analyzed using the 12-parameter regression prediction model developed by Dr. Steele (personal communication, Charles Steele, Ph.D.).

Ulnar BMC and BMD

Bone mineral measurements were determined by dual energy X-ray absorptiometry (DXA; QDR-4500A, Hologic, Inc., Bedford, MA, U.S.A) taken on the total ulna bone. From the DXA scan, BMC and BMD of the total ulna was determined using version 8.25a of the forearm software. Also, total body scans were performed in order to estimate arm fat-free mass (kg), arm fat mass (kg), and percent fat of the arm (%). Subjects were asked to enter the lab rested and in a relaxed state. Subjects were placed in the supine position on the DXA table and asked to remain still

until the scan was completed. Each total ulna scan comprised about two minutes.

Statistical Analysis

The main outcomes of this study assess whether or not there is a change in EI, arm strength, ulnar BMC, and ulnar BMD with training. Therefore, strength, EI, BMC, and BMD values were each evaluated across two treatments: time and training group, using a two-way repeated measures ANOVA to determine if each of these variables changed significantly with time in either isokinetic resistance training group, and/or between the trained and untrained limbs. Post-hoc testing was not performed as there were less than three groups. Secondary outcomes involve the determination of the ability of the change in arm peak torque, BMC, BMD, or fat-free mass of the arm to predict the change in ulnar EI. Change in total work of the arm, peak torque of the arm, BMD, BMC, and fat-free mass for each subject was calculated and entered in a linear regression model to determine the ability of each of these variables to explain the variability in ulnar EI. Statistical significance was set at 0.05. All data analysis was performed using SPSS (Version 11.0, 2001, SPSS Inc., Chicago, IL, U.S.A.).

RESULTS

The subjects in the final analysis comprised 49 college-aged females between the ages of 18 and 25 years. Table 3 displays the physical characteristics of the women included in the study. Subjects for

the current study achieved 80.2% adherence for the duration of the 32 week training protocol and reported no additional exercise outside the scope of our laboratory for the 32 week study period. There were no significant differences between groups at baseline for any variable assessed.

Strength Training.

Total Work. Analysis of all subjects regardless of training mode revealed no significant differences in the total work of either the trained or untrained arm when assessed for all subjects regardless of training mode. When assessed via training mode, neither the concentric group nor the eccentric group demonstrated a significant change in either the trained or untrained arm from baseline to 32 weeks. No time by training interaction could be determined for either the trained or untrained arm.

Peak Torque. Analysis of all subjects demonstrated a significant increase in both the trained (18% ↑, $P<0.01$) and untrained (7% ↑, $P=0.01$) arm throughout the 32 weeks of training. When analyzed across training mode, the eccentrically trained group displayed a significant increase in the trained limb (24% ↑, $P<0.01$) and a trend towards significance in the untrained limb ($P=0.06$) while the concentrically trained group showed a significant increase in the trained arm (14% ↑, $P<0.01$) with a non-significant increase in the untrained arm (Figure 4). No interaction was found in either the trained or untrained arm.

Ulnar Bending Stiffness. When all subjects were analyzed independent of training mode, ulnar bending stiffness displayed a significant increase in the trained arm (22% ↑, P=0.01) with no significant difference from baseline in the untrained arm (P=0.10) (Figure 5). When all subjects were assessed by training mode, those subjects in the eccentric group displayed a significant increase in the trained arm (40% ↑, P=0.01) with no significant increase in the untrained limb (Figure 6) while those who trained concentrically showed no significant change in either the trained or untrained arm (Figure 7). There was no time by training interaction.

Total body BMC and BMD. Neither total body BMC (P=0.5) nor total body BMD (P=0.7) was found to be significantly different from baseline (BMC = 2136.4 ± 277.7 g/cm, BMD = 1.11 ± 0.1 g/cm^2) values when global analysis was performed across all subjects. When subjects were analyzed according to training mode, no significant difference was found in either the concentric or eccentric group for BMC or BMD in either the trained or untrained limb. Also, no time by training interaction was determined in either BMC or BMD.

Total Ulna BMC and BMD.

BMC. Analysis of all subjects produced significant increases in the trained (4% ↑, P<0.01) and untrained arm (1% ↑, P<0.01) from baseline. When subjects were analyzed by training mode, the eccentrically trained group demonstrated significant increases in both the trained (2% ↑, P<0.01) and untrained (2% ↑, P<0.01) arm while the concentrically

trained group produced significant increases in the trained arm (2% ↑, P<0.01) with no increase in the untrained arm. However, there was no interaction effect found for time by training mode.

BMD. Analysis of the total subject pool demonstrated a significant increase in both the trained arm (2% ↑, P<0.01) and untrained arm (1% ↑, P<0.01) across all subjects. When assessed by training mode, the eccentric training group demonstrated significant increases in the trained (P<0.01, 2%) and untrained (2% ↑, P=0.03) limbs (Figure 8). Likewise, the concentrically trained group demonstrated significant increases throughout the training period in both the trained (3% ↑, P<0.01) and untrained (1% ↑, P<0.01) arm. However, no time by training interaction could be determined in the trained or untrained arm.

Fat-Free Mass of the Arm. Analysis of all subjects regardless of training mode revealed a significant increase in FFM in the trained arm (3% ↑, P<0.01) with no significant increase in the untrained arm. When assessed by training mode, the eccentric group demonstrated a significant increase in the trained arm (5% ↑, P<0.01) with no increase in the untrained arm while the concentric group showed no significant increase in either the trained or untrained limb. However, there was a significant time by training interaction in the trained arm (P=0.04) with no interaction in the untrained arm. Consequently, fat mass of the arm did not change throughout the duration of the study.

DISCUSSION

This study was conducted to determine if high-intensity muscular strength training would elicit an increase in mechanical strength of the ulna. Myburgh and colleagues have assessed a portion of this question by correlating activity level to ulnar bending stiffness; however, no training protocol was assigned (11). The present study is unique as it is the only longitudinal evaluation of ulnar EI. Previous studies performed by Myburgh et al, 1993; Kiebzak et al, 1999; and McCabe et al, 1991 were all cross-sectional and did not account for a training protocol which may have influence on bone stiffness values. Also, the current study is aimed at females aged 18-25 yrs. This is of importance because bone mineralization decreases with age; therefore, increases in bone stiffness imposed at youth may influence bone values in later life. Myburgh and colleagues (1993) employed 51 males aged 28-61; Kiebzak and colleagues (1999) recruited 125 women and classified then in age groups: ≤ 40 yr, 41-60 yr, and ≥ 61 yr; McCabe and colleagues (1991) divided two groups of women into a younger group (21-30 yr) and an older group (58-80 yr) (11,12,13). No studies to date have assessed the effect of high-intensity resistance training on ulnar bending stiffness. Also, the MRTA has undergone improvements since it was first designed which should enhance measurement validity and reliability. These early studies have relied on the first models developed to assess ulnar EI; however, refinements have been made through the work of Dr. Charles Steele, Ph.D. (personal communication) as well as Roberts and colleagues (1). Dr.

Steele has developed the 9- and 12-parameter models which may be a more representative model of how bone behaves *in vivo* as the proximal and distal ends of the bone are included in the final calculation of EI. Myburgh and colleagues employed the seven parameter model in their assessment of ulnar EI; however, Roberts and colleagues have found this method to be less than optimal for EI assessment of the tibia (1). Roberts and colleagues have shown that the algorithm for assessing tibial EI did not correlate well to three point bending experiments (1). Also, the 6-parameter model, which was a refinement of the 7-parameter model, assumes the bone is fixed at both ends. However, this is not how bone behaves *in vivo*. These distinct differences among previously reported literature between the training protocol chosen and the EI prediction model employed are of interest for two reasons: loading of the bone increases bone mineralization and technical advancements made to the MRTA device improves measurement validity and reliability.

We demonstrated that global analysis of all subjects revealed a 22% increase in ulnar EI with high-intensity resistance training. However, when subjects were analyzed according to training mode, the increase in ulnar EI within eccentric subjects was 40 % with no significant increase in the concentrically trained group. Therefore, this increase must be limited only to eccentric training rather than concentric training. In fact, eccentric training also produced significant increases in peak torque of the arm, total ulna BMC and BMD, and fat-free mass of the arm. The

response in the eccentrically trained arm was expected to change more so than the concentrically trained arm since it is known that eccentric training produces roughly 10-15 % greater force output than concentric training (14). However, we have shown here that eccentric training has the potential to yield results up to 40 % greater than concentric training. This is of importance as it may provide further insight into the loading stimulus necessary to promote macro- and micro-geometry changes in bone that are needed in increase mechanical strength. This study effectively demonstrates that high-intensity training does have a beneficial effect on increasing arm strength and bone mineral values and thus, may have a positive impact on increasing ulnar EI and decreasing fracture risk in young females. However, this may hold true only if a certain critical limit is achieved through training. The ulnar EI for the trained arm in the eccentric group increased from 16.0 ± 8.4 to 22.2 ± 10.2 at the end of the 32 week training protocol while the concentric group demonstrated a less pronounced effect increasing from 20.8 ± 10.1 to 23.4 ± 9.0. Multiple regression failed to demonstrate a systematic relation between the improvements in EI and changes in specific training variables that ostensibly should closely relate to the stimulus (e.g., muscle mass and muscular strength increases).

One potential confounder associated with our analysis for ulnar EI measurements was the low reliability of ulnar EI measurements. Steele and colleagues have indicated that ulnar EI measurements have

demonstrated reliability values in the range of 5.3% (8). However, in our lab using our version of the MRTA, reliability studies have been conducted in which variation can range from 27 - 59% between measurements of the tibia (15,16,17). It is important to note that ulnar EI and tibial EI are two very different measurements; however, the central theme is clear. Therefore, it is reasonable to assume that with standard deviations close to the mean, the reliability of the EI measurement was indeed poor. To assess this in our lab, a colleague recently proposed two possible mechanisms which could negatively affect the reliability within measurements: 1) positioning of the probe on the bone and 2) subject movement once the probe is positioned (18). Only slight modifications in these two variables may have tremendous implications in the measurement of ulnar EI.

With respect to the amount of work performed during training, we found that total work of the arm did not differ significantly from baseline to post training when assessed across all subjects regardless of training mode. This may be attributed to the specific protocol chosen. A high intensity training regimen was followed by these subjects and this produced significant increases in peak torque. This may be due to the fact this was indeed a strength training protocol and a lesser effect was seen on total work because the protocol was not designed to increase muscular endurance. Also, there several main factors beyond the study parameters which may also play a role in a non-significant increase in

total work of the arm. First, the subjects in the present study were all untrained females with little to no prior experience with resistance training. We proposed that high-intensity resistance training may induce greater responses in bone stiffness compared to normal physical activity; however, these women may have been become too accustomed to the training protocol due to the length of the study. It is known that in order to produce maximal gains in strength, the exercise must remain novel in order to elicit the proper response. Therefore, the duration of the training period may have been too long and training effects should have been assessed at an earlier date. Also, subject motivation may have played a role in the willingness of subjects to work at the required level of effort. We asked these women to maintain 90-100 % effort on every repetition they performed; however, some of these women did not train to the level we required either due to their self reported lack of sleep, boredom with the protocol, or sessions missed. Most of these women admitted to less than five hours of sleep previous to entering our lab for training. Meney and colleagues found that after one night of sleep deprivation, muscular strength decreased (19). Also, these women admitted a tendency to become accustomed to the training protocol and did not give the effort required to further strength gains. Therefore, the exercise protocol may not have been novel enough to attain strength gains over 32 weeks. Also, since all of these subjects were college students, training sessions continued through fall, winter, and spring breaks. We assigned these

women an alternative workout in order to train away from our lab; however, we have no definitive means of assessing whether these training sessions took place. Although the compliance rate was 80.2%, most of these women did miss at least one session per week which usually translated into about four days with no training if the session missed was a Friday or Saturday.

Although alterations in total body bone mineral values were not determined to be significant, total ulna BMC and BMD values were significant from baseline to post training in the trained arm in the eccentric and concentric groups. The percent increase in any of these groups did not exceed 3%; however, this small increase may have the ability to produce large increases in ulnar EI. Since the only training these women received was localized to the upper arm, this stimulus was probably not adequate to elicit a total body response to increase bone mineral values. However, only small increases in new bone material are necessary to impact large increases in the integrity of bone. This is due to bone material being added to areas to which the greatest stresses are applied and in this case, to the upper arm regions. With this in mind, bone can become less likely to fracture through only small increases in the amount of bone material added as seen with a 40% increase in ulnar EI in those who trained eccentrically.

This study was aimed at determining if high-intensity resistance training could improve ulnar bending stiffness in young females.

Although this effect did not occur in the concentric group, a 40% increase was seen in the eccentrically trained group. Also, small but significant increases in ulnar bone mineral values were seen in both the concentrically and eccentrically trained groups. This may be an indication for future studies in assessing the effects of high-intensity resistance training in eliciting a positive effect on bone health. For future studies, the training protocol should encompass varied resistance throughout the training program with different exercises performed and the resistance gradually increased. Most importantly, this training protocol must employ eccentric contractions to enhance the effect of training on bone health. This may effectively alleviate the problems associated with a same routine exercise program in evoking a maximal muscular response. Also, future studies should look at this protocol at the long term level. The same variables measured in the current study should be assessed at later time points to assess if these small increases in bone mineral values will contribute to an increased ulnar bending stiffness from months to years.

Our findings have shown that eccentric training is superior to concentric loading for eliciting the proper threshold limit required for bone adaptation. Since concentric training did not indicate a relationship between the change in strength variables and the change in ulnar EI, it may be assumed that this mode of training does not reach the threshold required for bone adaptation, even though this mode of training may

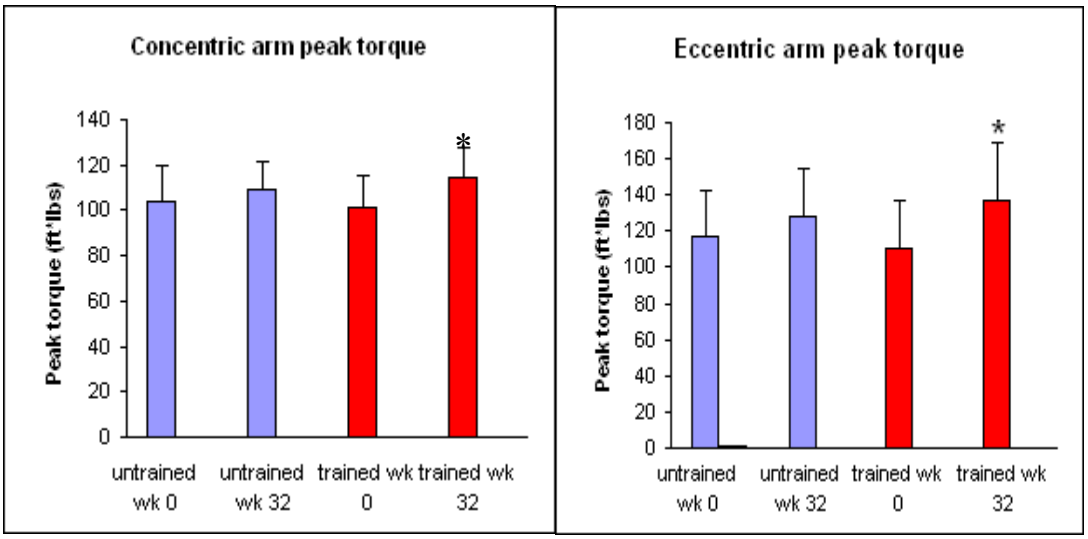
impact bone mineral values. These data are in support for a threshold hypothesis, in which changes to bone geometry will not occur unless a threshold limit has been reached. However, many aspects associated with our version of the MRTA may have contributed to the lack of findings within the concentric group. For further analysis to be performed, it is necessary that the MRTA undergo improvements in order to accurately determine the bending stiffness of long bones *in vivo*.

Tables and Figures

Table 3.
Physical Characteristics of subjects (means ± SD).

N	Age (yrs)	Height (cm)	Weight (kg)	BMI (kg/m²)	Nondominant EI (Nm²)	Dominant EI (Nm²)
49	19.9 ± 1.7	163.3 ± 6.1	59.7 ± 8.7	22.5 ± 3.0	18.8 ± 9.7	19.7 ± 7.2

N = number of subjects



*=P<0.05
Figure 4. Concentric and Eccentric Peak Torque.

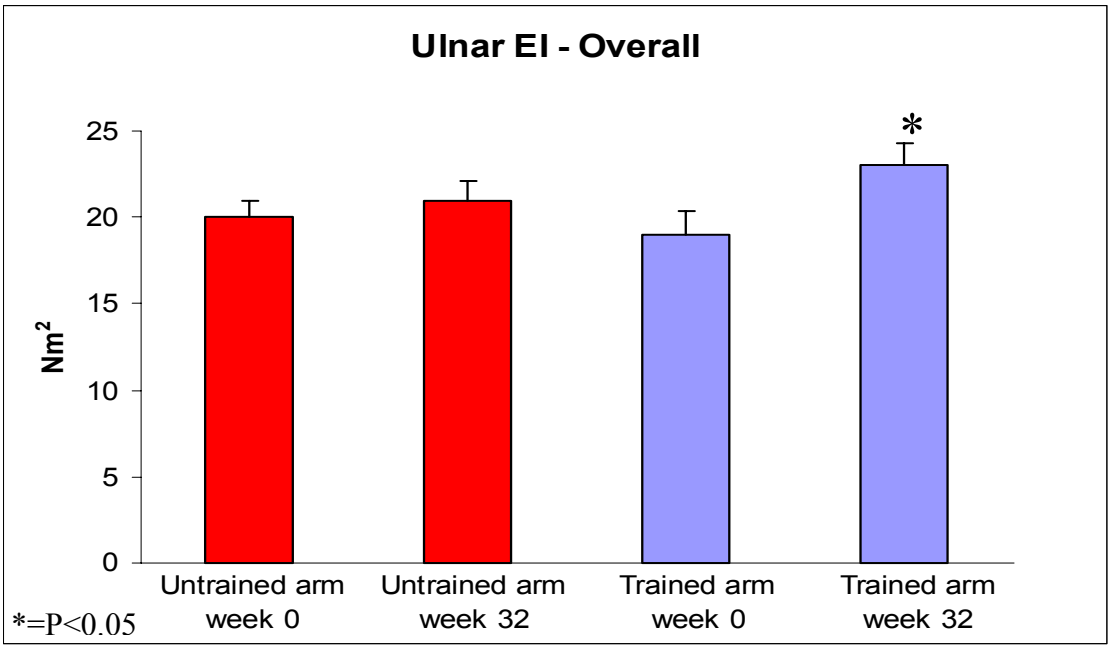


Figure 5. Global Ulnar EI for all subjects.

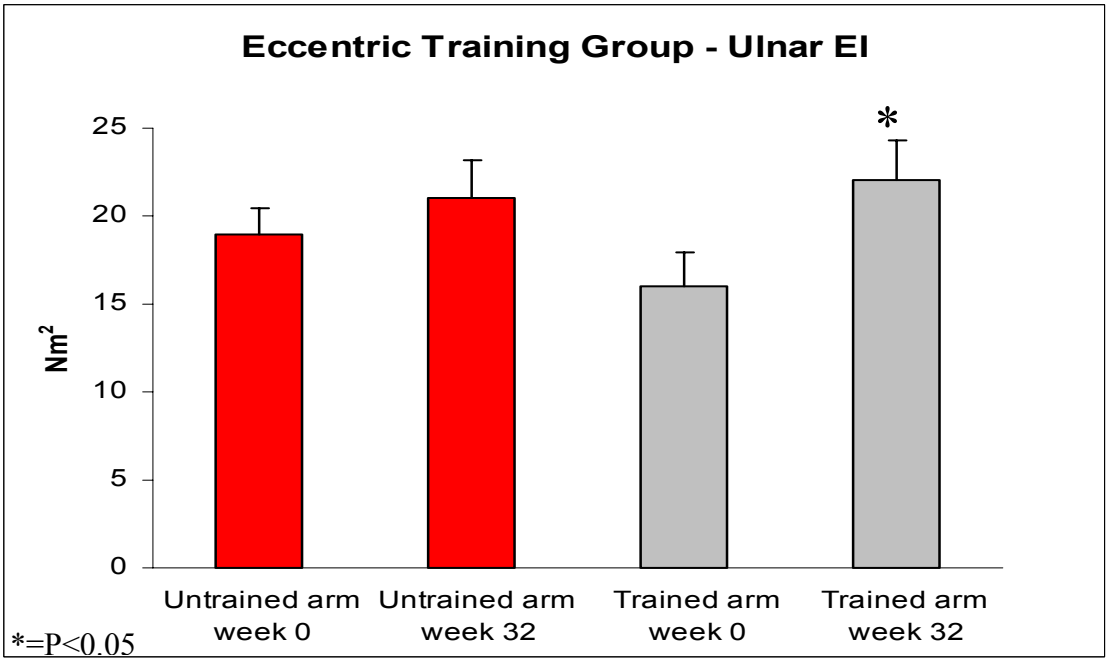


Figure 6. Eccentric Ulnar EI in the trained and untrained arm.

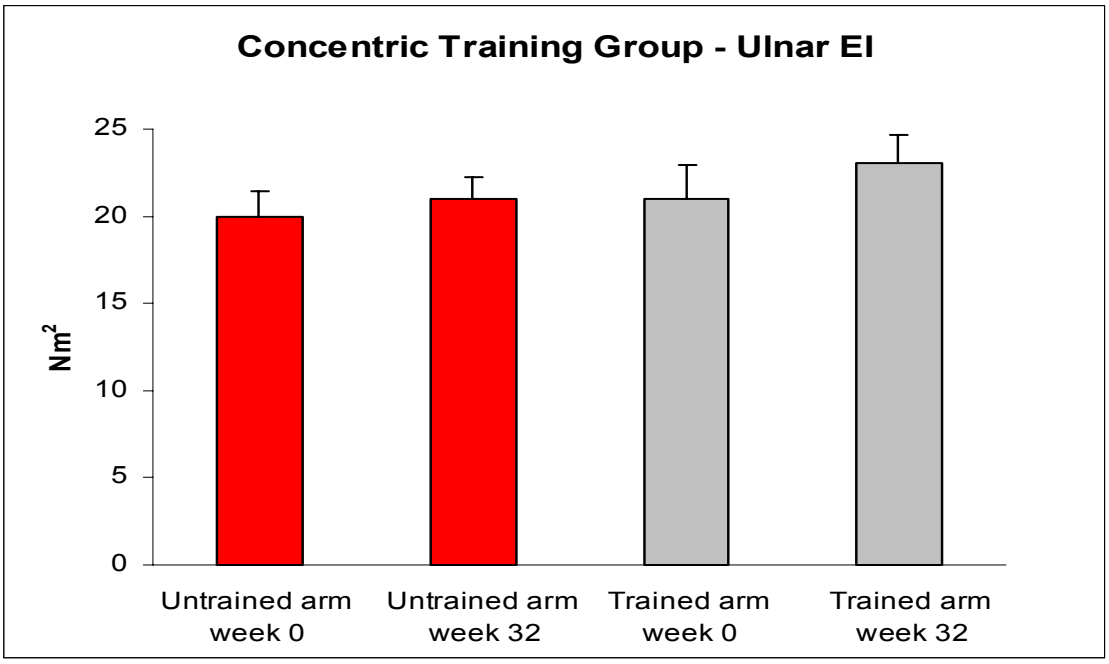
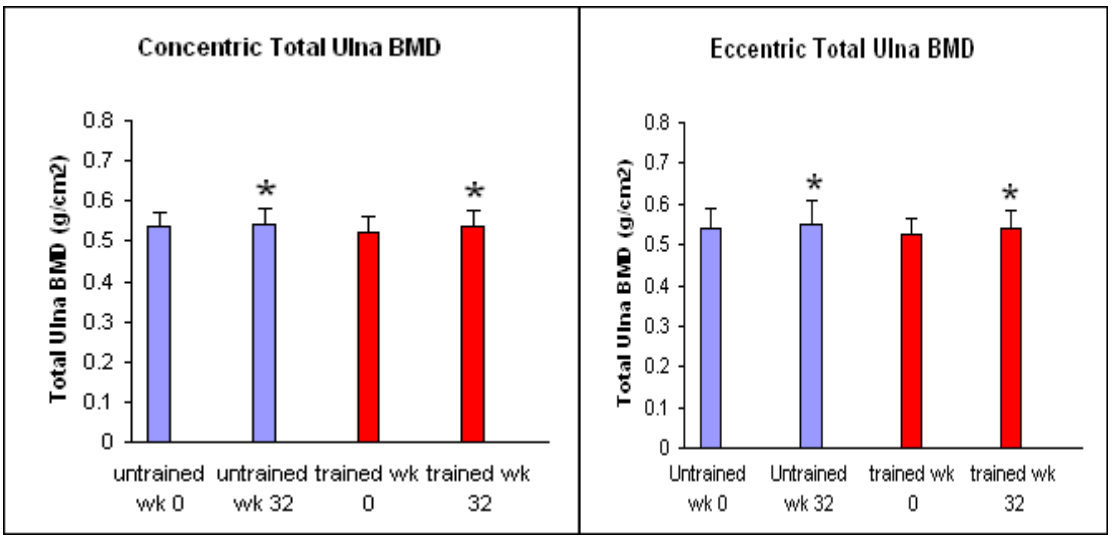


Figure 7. Concentric Ulnar EI in the trained and untrained arm.



*=P<0.05

Figure 8. Total Ulna BMD for Concentric and Eccentric Groups.

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Chapter IV

Summary, Implications for clinical practice and research

The purpose for the study was to determine the effects of high-intensity resistance training on ulnar bending stiffness in young college-aged females. Osteoporotic fractures, especially in women, are of increasing importance to our society as one in six white women is predicted to suffer a hip fracture sometime in their life (Cummings et al, 1995).

In the present study, although ulnar EI was found only to reach significance from baseline to post-training in the eccentrically trained group, this effectively demonstrates that high-intensity resistance training has potential implications for increasing bone stiffness in young women. As a consequence, strength training does possess implications for possibly decreasing the risk of fragility fractures in women. In fact, total ulnar BMC and BMD were both increased significantly throughout the duration of the study indicating bone material was indeed added to the bone as a result of stressing the bone by training. Of course, this is of significance due to the geometry and mineralization of bone being altered for the better. These two properties influence the strength of bone and therefore, should increase ulnar EI even if only small amounts of bone mineral are added through a response in resistance training.

Recommendations for further research

Through the findings in the current study and those performed by others, the following recommendations are necessary to further enhance the usage of the MRTA to assess fracture risk:

1) Improvements need to be made within in the MRTA device itself before further analysis can be performed on long bone bending stiffness. In order to obtain valid and reliable results, the MRTA should be modified to incorporate operator failsafe procedures. As of now, operator error is a major limitation in obtaining reliable data. Future studies should include a method whereby the operator may view the MRTA curve, as well as the RMS limit criterion, to ensure a proper measurement was obtained before performing additional tests. Also, the MRTA needs to be fitted with a device that will allow the limb beings assessed to remain in a fixed state thereby eliminating the chance of movement during a single test or between subsequent tests. Once these modifications have been produced, the MRTA should be able to estimate bending stiffness in a very reliable manner.
2) The algorithm in which bending stiffness is assessed needs to be validated. Previous studies have validated the 6- and 7-parameter models (Roberts et al., 1996 and Steele et al, 1988) but these do not present an actual representation on how bone behaves *in vivo*. Steele has developed a closer representation on how bone behaves *in vivo*, the 9- and 12-parameter models, by the addition of accounting for stiffness, mass, and damping at the distal and proximal ends of bone.

This involves a testing arrangement in which neither bone end is restricted or tightly clamped, as in the latest modification of the MRTA apparatus constructed for this lab by Charles Steele (personal communication, Dr. Charles Steele, Ph.D.). However, no validation studies on the 9- or 12-parameter model have been performed to date.

3) Strength training is known to produce increases in bone mass. However, there are many different methods of strength training. Our study focused on isokinetic strength training with either concentric or eccentric contractions. Future studies should incorporate many different modes of strength and endurance training to determine the effect on bending stiffness in long bones. For instance, group A could train using isokinetic resistance, group B could train with isometric resistance, while group C trains using isotonic resistance. After a defined training period, bending stiffness can be assessed to determine which training method may offer the greatest benefits.

Recommendations for clinical applications

1) Once the MRTA has been validated for measurement reliability and the 12- parameter algorithm, indices of bending stiffness values should be constructed in which fracture risk prediction can be assessed. As of now, osteoporosis is clinically diagnosed by DXA technology; thereby fracture prediction risks can be obtained by the relative BMC. However, since BMC does not explain bone strength in its entirety,

this fracture prediction does have limits. Since the MRTA is able to estimate the two properties affecting bone strength, it is reasonable to assume that fracture prediction using the MRTA would produce enhanced fracture prediction risks above DXA. Once the initial concerns are alleviated within the MRTA device, future studies should focus on producing a nomogram to determine an individual's relative risk of fracture based on EI_{MRTA} results and combine this with DXA results for a more robust predictor of fracture occurrence.

2) Studies should be conducted using the MRTA in a clinical setting. The MRTA offers several advantages compared to DXA in that the MRTA is relatively inexpensive, easy to use, and much more portable that the DXA device. This technology could be applied in traveling heath programs such as those which assess blood lipid levels. This has the potential for great demand, especially in rural areas where individuals may have no access to a DXA.

3) Since many fractures that occur in women involve the hip, the MRTA could be used to assess the bending stiffness of smaller, non-appendecular bones. Future studies could assess these areas where fractures occur, mainly late in age and in osteoporotic women. Once these areas are assessed, individuals would be better informed of their relative risk of these types of fractures while they are young so that intervention could be applied.

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Appendices

APPENDIX A

Informed Consent

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY
Informed Consent for Participants of Investigative Project

Title of Study: Effects of Isokinetic Eccentric Muscular Strength Training of the Arm on Ulnar Bending Stiffness in Young Women Aged 18-25.

Location of Study: War Memorial Hall, Virginia Polytechnic and State University, Blacksburg VA

Principal Investigators: William G. Herbert, Ph.D., Shelly Nickols-Richardson, Ph.D., Warren Ramp, Ph.D.

Informed Consent:

I understand that any information I provide will be confidential. Information provided on this form will only be used by the investigators to determine my initial eligibility for inclusion into the research study, "Effects of isokinetic resistance training and deconditioning on bone stiffness, bone mineral density, and bone turnover in military-aged women." I understand that by completing this form, I am not guaranteed inclusion into the study. I also understand that information provided on this form may be used by the investigators to contact me.

Please complete the following form. We will use this information to determine your eligibility for participation in the study. A member of the research team will contact you regarding your eligibility status.

Please note: Any information passing from your computer to the server is encrypted for security purposes.

Please provide the following contact information:	
First Name:	

Last Name:	
Street Address:	
City:	
State/Province:	
Zip/Postal Code:	<>
Country:	
Work Phone:	
Home Phone:	
Fax:	
E-mail address:	

Please complete:	
Date of birth:	/ /
Sex:	Female Male
Height:	feet inches
Weight:	lbs.

Please answer the following questions:	
(1) Do you have any known heart, lung, or metabolic disease (i.e., diabetes)?	Yes No
(2) Do you have any known musculoskeletal condition that would prevent you from participating in strenuous exercise?	Yes No
(3) Have you had any bone fractures within the past year?	Yes

	☐ No
(4) Have your menstrual cycles been regular (more than 8 times) within the past year?	☐ Yes ☐ No
(5) Are you pregnant?	☐ Yes ☐ No
(6) Are you presently using contraceptives?	☐ Yes ☐ No
If so, please indicate which type you are using below: [ ]<>	
(7) Are you, or will you be a student at an area college during the 2000-2001 academic year? ***Note:* You do <u>not</u> have to be a student to participate in this project.**	☐ Yes ☐ No
If so, please indicate your class standing as of Fall 2001: ☐ Freshman ☐ Sophomore ☐ Junior ☐ Senior ☐ Graduate Student	
(8) Please enter the date of your last physical exam *(MM/DD/YY)*:	[ ] / [ ] / [ ]
Clear Submit	

APPENDIX B

Medical/Health History Questionnaire

VIRGINIA TECH
LABORATORY FOR HEALTH AND EXERCISE SCIENCE
Medical and Health History Form
Title of Project: Effect of isokinetic resistance training on ulnar stiffness

in young, college-aged women

Name: Age: _____Date of Birth: _____
Local
Address:_____
_____
Telephone Number: _____
Email Address: _____
Address for Permanent Residence:
_____

Medical History
Please indicate any current or previous conditions or problems you have experienced or have been told by a
physician you have had:

	Yes	**No**
Circulation problems:	_____	_____
Kidney disease or problems:	_____	_____
Musculoskeletal problems:	_____	_____
Broken bones (Past 12 months)	_____	_____
High blood pressure:	_____	_____
High blood cholesterol:	_____	_____
Diabetes:	_____	_____
Eating disorders (bulimia, anorexia):	_____	_____

If "yes" to any of the above please indicate the date, explain, and describe:

Please list any hospitalizations/operations/recent illnesses (Type/Date):

Health Habits
Are you on any special type of diet? Yes ___ No ___
If "yes" please describe

Do you drink alcoholic beverages? Yes _____No _____How many drinks per week? _____
Do you smoke cigarettes? Yes _____ No _____ Packs per day: _____

Exercise Habits
Do you engage in regular exercise (more than one session/week)? Yes _____ No _____
If "yes" please list:

Activity Frequency (times per week) Duration (minutes)

Explanation of exercise activities:
_____
_____
_____
_____
_____

Questions Related to Reproductive Function
Do you use birth control? Yes _____ No _____
If "yes" what form of birth control:
Date of last menses:
Have you had any abnormal menses or absence of menses in the last 12 months? Yes _____ No _____
If .yes., describe this menstrual problem:
_____
_____
_____
Please list all medications (prescription and over-the-counter) you are currently taking or have taken in the
past week:

Please sign to indicate the above information is correct:

_____ _____
Print Name Signature Date

APPENDIX C

Tables and Figures

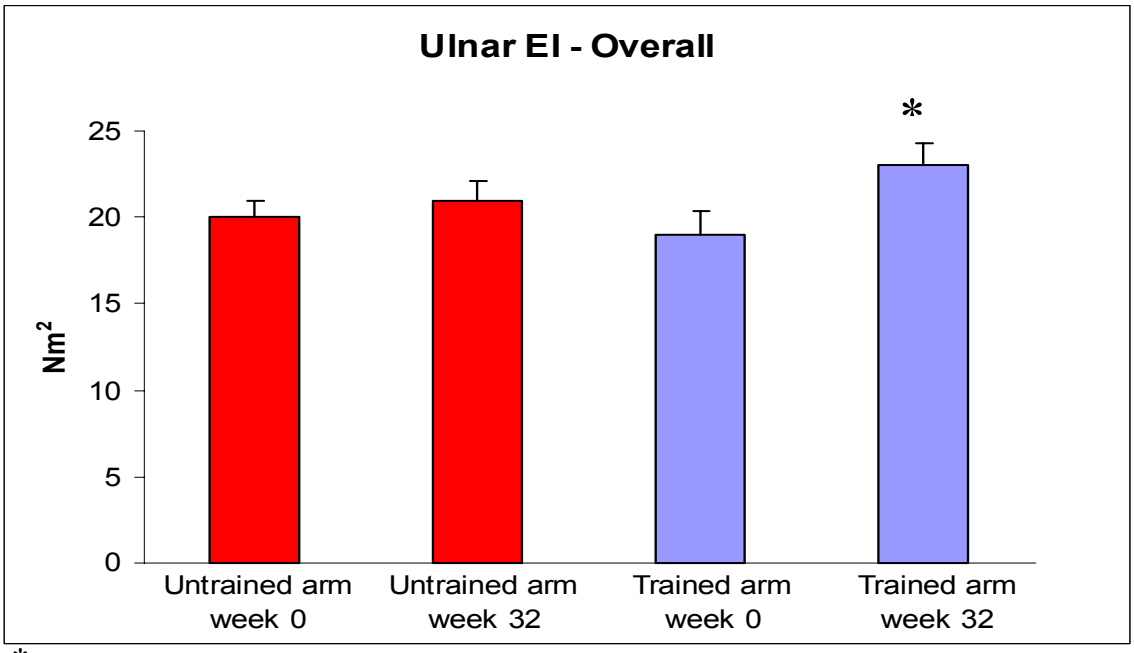
Table 1.
12-parameter model for assessing ulnar EI.

Bone	Soft Tissue	Proximal Bone	Distal Bone
Mass	Mass	Mass	Mass
Stiffness	Stiffness	Stiffness	Stiffness
Damping	Damping	Damping	Damping

Table 2.
Physical Characteristics of subjects (means ± SD).

N	Age (yrs)	Height (cm)	Weight (kg)	BMI (kg/m²)	Nondominant EI (Nm²)	Dominant EI (Nm²)
49	19.9 ± 1.7	163.3 ± 6.1	59.7 ± 8.7	22.5 ± 3.0	18.8 ± 9.7	19.7 ± 7.2

N = number of subjects



* = P < 0.05

Figure 1. Global Ulnar EI for all subjects.

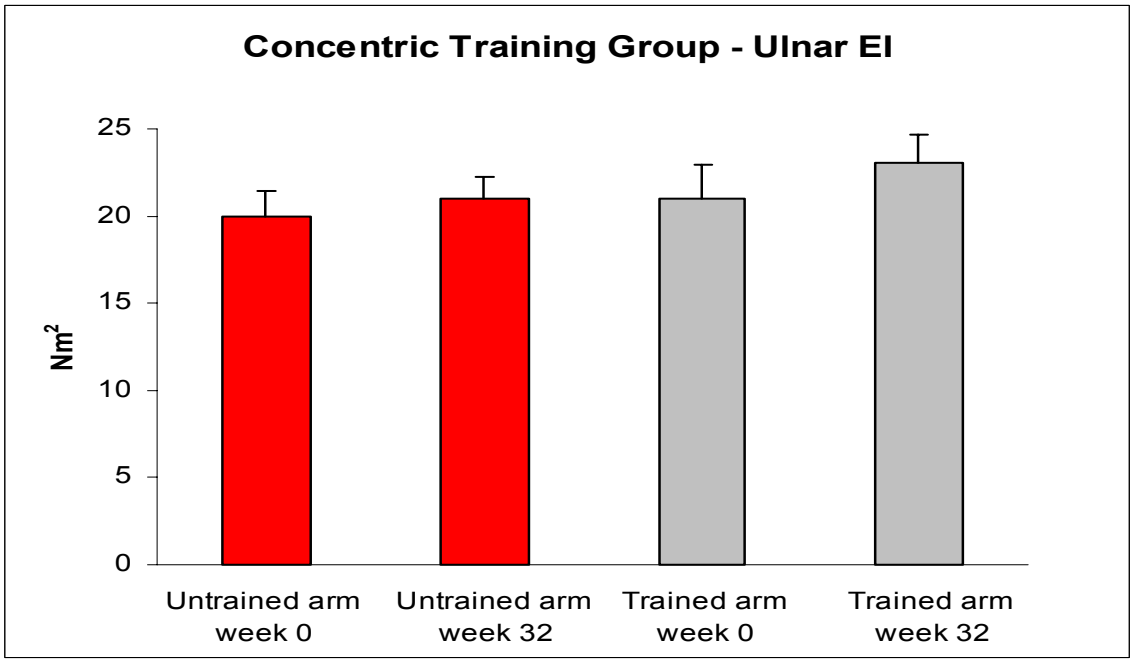
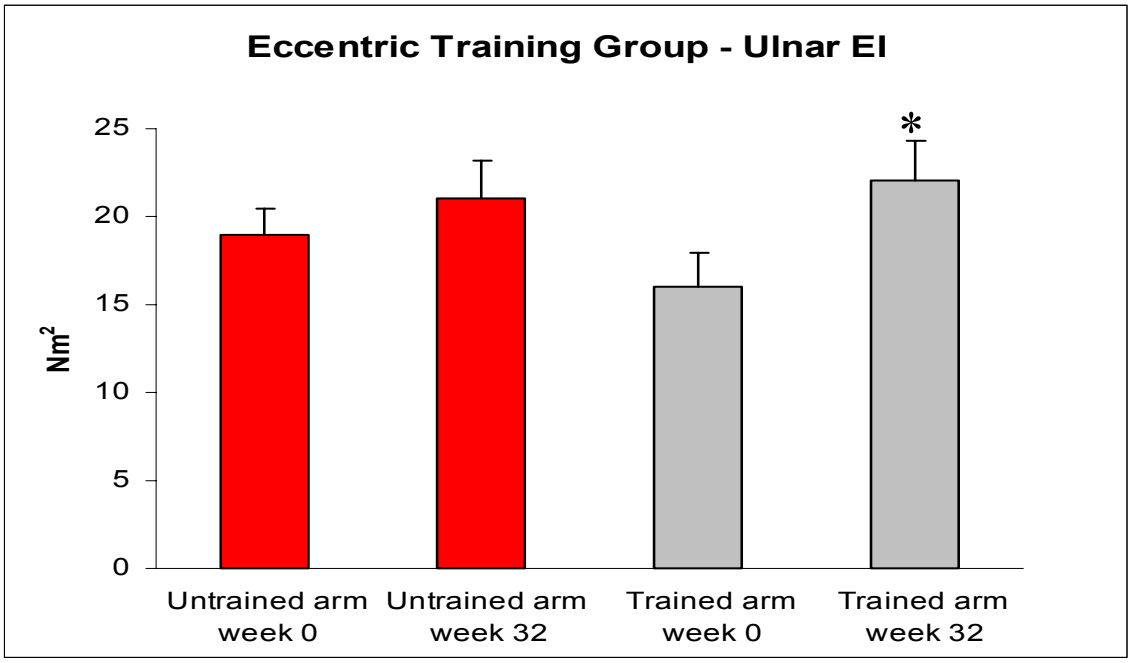


Figure 2. Concentric Ulnar EI in the trained and untrained arm



* = P < 0.05

Figure 3. Eccentric Ulnar EI in the trained and untrained arm.

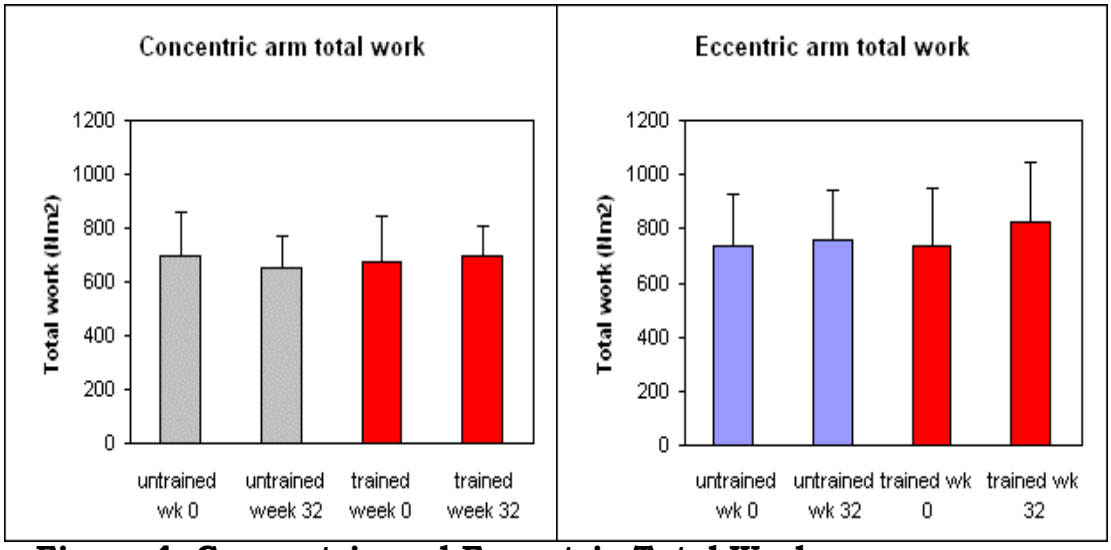
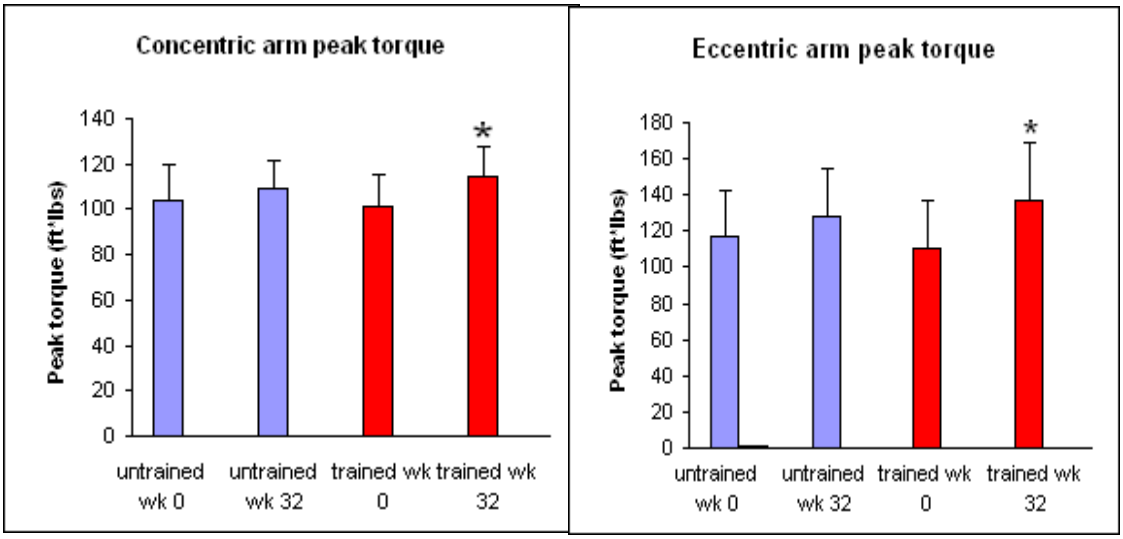
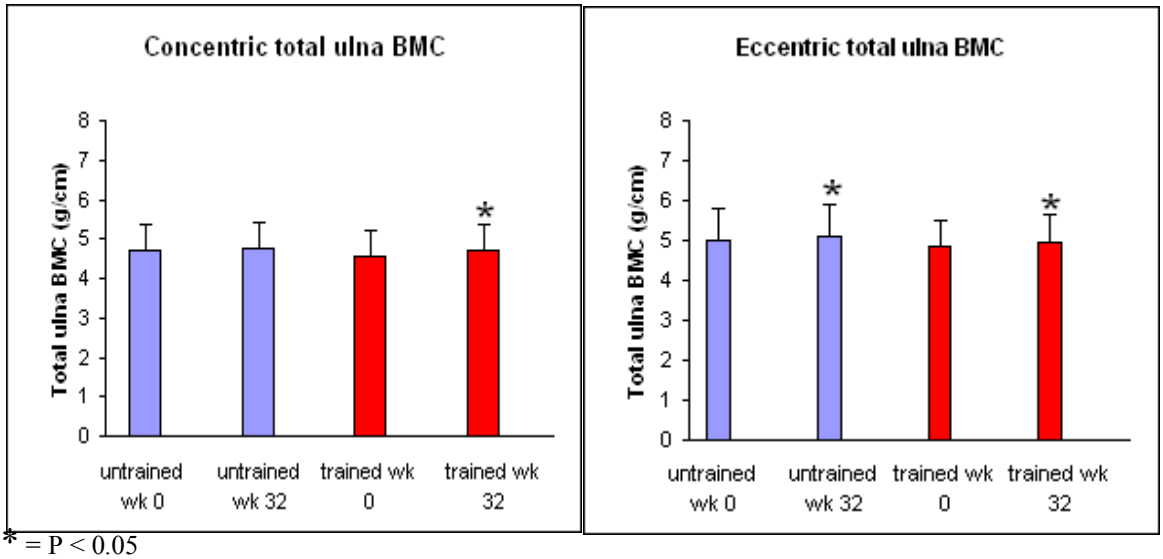


Figure 4. Concentric and Eccentric Total Work.



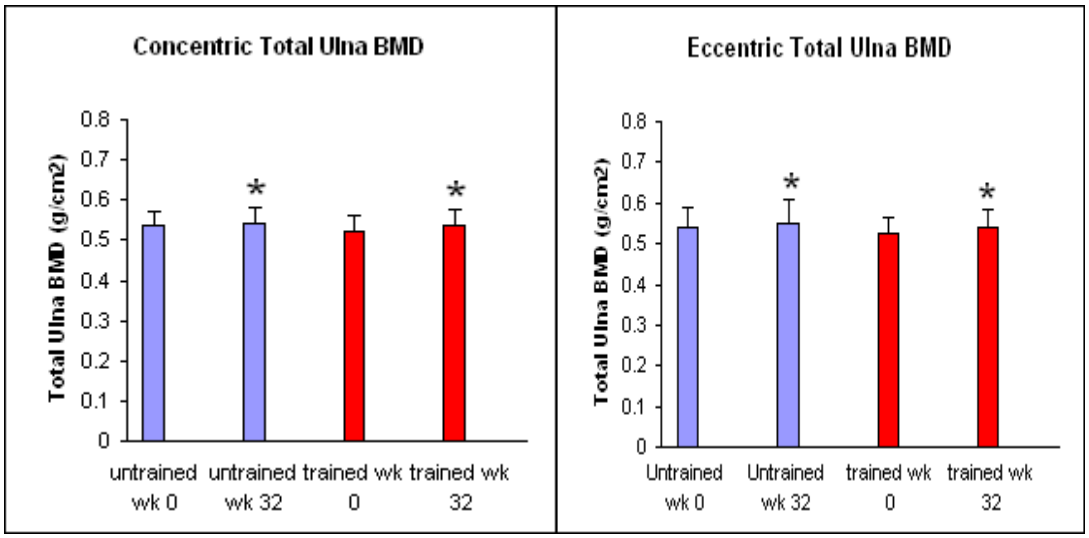
* = P < 0.05

Figure 5. Concentric and Eccentric Peak Torque.



* = P < 0.05

Figure 6. Total Ulna BMC for Concentric and Eccentric Groups.



* = P < 0.05

Figure 7. Total Ulna BMD for Concentric and Eccentric Groups

APPENDIX D

Summary ANOVA tables

Ulnar Bending Stiffness

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	408.793	1	408.79	7.27	0.01
	UNTR	69.557	1	69.56	2.72	0.11
Error(TIME)	TR	2699.543	48	56.24		
	UNTR	1226.480	48	25.55		

All subjects by training mode

CONCENTRIC

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	98.266	1	98.27	1.74	0.20
	UNTR	33.607	1	33.61	1.37	0.25
Error(TIME)	TR	1578.566	28	56.38		
	UNTR	686.369	28	24.51		

ECCENTRIC

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	388.502	1	388.50	7.08	0.02
	UNTR	36.888	1	36.89	1.30	0.27
Error(TIME)	TR	1043.002	19	54.89		
	UNTR	539.173	19	28.38		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	462.102	1	462.10	8.28	0.01
	UNTR	70.160	1	70.16	2.69	0.11
TIME * EX_GROUP	TR	77.975	1	77.97	1.40	0.24
	UNTR	0.938	1	0.94	0.04	0.85
Error(TIME)	TR	2621.568	47	55.78		

UNTR	1225.542	47	26.08	

Total body BMC

All subjects

Tests of Within-Subjects Effects

Measure: BMC

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	1394.373	1	1394.37	0.40	0.53
Error(TIME)	169166.441	48	3524.30		

All subjects by training mode

Concentric

Tests of Within-Subjects Effects

Measure: BMC

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	1030.825	1	1030.83	0.27	0.61
Error(TIME)	105908.687	28	3782.45		

Eccentric

Tests of Within-Subjects Effects

Measure: BMC

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	391.528	1	391.53	0.12	0.74
Error(TIME)	63229.773	19	3327.88		

Group by time interaction

Tests of Within-Subjects Effects

Measure: BMC

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	1276.951	1	1276.95	0.35	0.55
TIME * EX_GROUP	27.981	1	27.98	0.01	0.93
Error(TIME)	169138.460	47	3598.69		

Total body BMD

All subjects

Tests of Within-Subjects Effects

Measure: BMD_TB

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	0.000	1	0.00	0.10	0.76
Error(TIME)	0.012	48	0.00		

All subjects by training mode

Concentric

Tests of Within-Subjects Effects

Measure: BMD_TB

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	0.000	1	0.00	0.00	0.99
Error(TIME)	0.009	28	0.00		

Eccentric

Tests of Within-Subjects Effects

Measure: BMD_TB

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	0.000	1	0.00	0.33	0.57
Error(TIME)	0.004	19	0.00		

Group by time interaction

Tests of Within-Subjects Effects

Measure: BMD_TB

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	0.000	1	0.00	0.14	0.71
TIME * EX_GROUP	0.000	1	0.00	0.15	0.70
Error(TIME)	0.012	47	0.00		

Total Ulna BMC

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.313	1	0.31	20.56	0.00
	UNTRAIN	0.101	1	0.10	9.57	0.00
Error(TIME)	TRAINED	0.731	48	0.02		
	UNTRAIN	0.509	48	0.01		

All subjects by training mode

Concentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.163	1	0.16	8.73	0.01
	UNTRAIN	0.028	1	0.03	2.31	0.14
Error(TIME)	TRAINED	0.521	28	0.02		
	UNTRAIN	0.344	28	0.01		

Eccentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.152	1	0.15	13.93	0.00

	UNTRAIN	0.087	1	0.09	11.04	0.00
Error(TIME)	TRAINED	0.208	19	0.01		
	UNTRAIN	0.150	19	0.01		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.311	1	0.31	20.06	0.00
	UNTRAIN	0.112	1	0.11	10.67	0.00
TIME * EX_GROUP	TRAINED	0.002	1	0.00	0.12	0.73
	UNTRAIN	0.014	1	0.01	1.36	0.25
Error(TIME)	TRAINED	0.729	47	0.02		
	UNTRAIN	0.494	47	0.01		

Total Ulna BMD

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.004	1	0.00	44.45	0.00
	UNTRAIN	0.002	1	0.00	12.16	0.00
Error(TIME)	TRAINED	0.004	48	0.00		
	UNTRAIN	0.006	48	0.00		

All subjects by training mode

Concentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.003	1	0.00	31.48	0.00
	UNTRAIN	0.000	1	0.00	9.12	0.01
Error(TIME)	TRAINED	0.002	28	0.00		
	UNTRAIN	0.002	28	0.00		

Eccentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.001	1	0.00	13.37	0.00
	UNTRAIN	0.001	1	0.00	5.24	0.03
Error(TIME)	TRAINED	0.002	19	0.00		
	UNTRAIN	0.005	19	0.00		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	0.004	1	0.00	41.11	0.00
	UNTRAIN	0.002	1	0.00	13.32	0.00

TIME * EX_GROUP	TRAINED	0.000	1	0.00	0.24	0.63
	UNTRAIN	0.000	1	0.00	1.32	0.26
Error(TIME)	TRAINED	0.004	47	0.00		
	UNTRAIN	0.006	47	0.00		

Total Work

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	59174.298	1	59174.30	2.60	0.11
	UNTR	7265.437	1	7265.44	0.41	0.52
Error(TIME)	TR	1094207.355	48	22795.99		
	UNTR	846592.757	48	17637.35		

All subjects by training mode

Concentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	5417.130	1	5417.13	0.36	0.55
	UNTR	25820.080	1	25820.08	1.91	0.18
Error(TIME)	TR	418293.189	28	14939.04		
	UNTR	378010.657	28	13500.38		

Eccentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	85340.502	1	85340.50	2.52	0.13
	UNTR	3608.893	1	3608.89	0.15	0.70
Error(TIME)	TR	644330.832	19	33912.15		
	UNTR	446418.563	19	23495.71		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TR	73854.101	1	73854.10	3.27	0.08
	UNTR	3185.831	1	3185.83	0.18	0.67
TIME * EX_GROUP	TR	31583.334	1	31583.33	1.40	0.24
	UNTR	22163.537	1	22163.54	1.26	0.27
Error(TIME)	TR	1062624.021	47	22609.02		
	UNTR	824429.220	47	17541.05		

Peak Torque

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	8813.465	1	8813.47	28.92	0.00
	UNTRAINE	1358.358	1	1358.36	6.85	0.01
Error(TIME)	TRAINED	14628.872	48	304.77		
	UNTRAINE	9519.248	48	198.32		

All subjects by training mode

Concentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	2713.959	1	2713.96	16.51	0.00
	UNTRAINE	371.990	1	371.99	2.87	0.10
Error(TIME)	TRAINED	4603.871	28	164.42		
	UNTRAINE	3628.084	28	129.57		

Eccentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	7092.005	1	7092.00	14.92	0.00
	UNTRAINE	1187.764	1	1187.76	3.97	0.06
Error(TIME)	TRAINED	9032.503	19	475.39		
	UNTRAINE	5689.768	19	299.46		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	TRAINED	9617.596	1	9617.60	33.15	0.00
	UNTRAINE	1508.194	1	1508.19	7.61	0.01
TIME * EX_GROUP	TRAINED	992.498	1	992.50	3.42	0.07
	UNTRAINE	201.396	1	201.40	1.02	0.32
Error(TIME)	TRAINED	13636.374	47	290.14		
	UNTRAINE	9317.852	47	198.25		

Fat-Free Mass of the Arm

All subjects

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	ARMFFMTR	67853.071	1	67853.07	10.25	0.00
	ARMFFMUN	1093.332	1	1093.33	0.12	0.73
Error(TIME)	ARMFFMTR	317794.916	48	6620.73		
	ARMFFMUN	421916.866	48	8789.93		

All subjects by training mode

Concentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	ARMFFMTR	9525.303	1	9525.30	1.10	0.30
	ARMFFMUN	669.669	1	669.67	0.06	0.80
Error(TIME)	ARMFFMTR	242709.075	28	8668.18		
	ARMFFMUN	295508.663	28	10553.88		

Eccentric

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	ARMFFMTR	84217.366	1	84217.37	32.53	0.00
	ARMFFMUN	424.137	1	424.14	0.06	0.80
Error(TIME)	ARMFFMTR	49196.243	19	2589.28		
	ARMFFMUN	126407.728	19	6653.04		

Group by time interaction

Univariate Tests

Source	Measure	Type III Sum of Squares	df	Mean Square	F	Sig.
TIME	ARMFFMTR	81572.021	1	81572.02	13.13	0.00
	ARMFFMUN	1048.234	1	1048.23	0.12	0.73
TIME * EX_GROUP	ARMFFMTR	25889.597	1	25889.60	4.17	0.05
	ARMFFMUN	0.475	1	0.48	0.00	0.99
Error(TIME)	ARMFFMTR	291905.319	47	6210.75		
	ARMFFMUN	421916.391	47	8976.94		

Multiple Regression Analysis-
Dominant Arm

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	0.25	0.06	-0.07	7.40

Predictors: (Constant), bmd difference, ffm dominant difference, body fat difference, peak torque dominant difference, total work dominant difference, bmc difference

Dependent Variable: ei dominant difference

ANOVA

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	151.877	6	25.31	0.46	0.83
	Residual	2301.083	42	54.79		
	Total	2452.960	48			

Predictors: (Constant), bmd difference, ffm dominant difference, body fat difference, peak torque dominant difference, total work dominant difference, bmc difference

Dependent Variable: ei dominant difference

Multiple Regression Analysis- Non-dominant Arm

Model Summary

Model	R	R Square	Adjusted R Square	Std. Error of the Estimate
1	0.27	0.07	-0.06	10.91

Predictors: (Constant), bmd difference, ffm nondominant difference, body fat difference, total work nondominant difference, peak torque nondominant difference, bmc difference

Dependent Variable: ei nondominant difference

ANOVA

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	397.329	6	66.22	0.56	0.76
	Residual	5001.756	42	119.09		
	Total	5399.085	48			

Predictors: (Constant), bmd difference, ffm nondominant difference, body fat difference, total work nondominant difference, peak torque nondominant difference, bmc difference

Dependent Variable: ei nondominant difference

APPENDIX E

Raw Data

Total Work of the arm (ft·lbs)

Subject	Total Work dominant arm week 0	Total Work dominant arm week 32	Total Work non-dominant arm week 0	Total Work non-dominant arm week 32
1	719.7	831.9	729.9	773.1
2	1081.7	948.4	1098.9	971.2
3	390.4	567	398.6	603.2
12	736.7	601.5	707.7	608.9
13	518.9	563.7	559	625.1
16	570.5	611.3	591.7	620
30	804	653.4	681	686.8
31	558.8	644.7	545.2	654.5
32	755.2	568	685	676.5
33	659.4	684.2	633.3	649.5
36	1059.5	516.4	1073.3	535.9
42	550.7	700.1	595.3	723.9
50	638.7	669.5	551.6	783.6
52	738.9	783.3	641.4	750.8
61	505.4	629.7	496.6	759.3
64	556.6	830.1	543	843.7
65	816.8	689.52	882.2	741.13
67	914.9	674.1	926.8	699.9
81	558.2	423.3	479.1	492.6
83	937.7	782.5	875.9	921.3
87	708.7	678.8	742.5	581.9
94	704.6	781.7	673.2	857.7
308	790.1	542.4	676.1	507.9
330	536.3	513.2	517.2	707.2
335	710.37	672.09	715.57	730.35
352	777.7	722	710	736.6
354	667.6	520.6	653	618
361	558.84	661.35	567.9	723.71
363	685	523.4	659.4	586.6
5	1280.5	839.01	1460.3	961.43
11	749.5	1075	860.5	1133.9
14	687.4	665.4	619.3	676
25	659.7	805.7	660.2	819.6
37	1171.3	941.6	1029.9	985.4
38	640	489.1	632.2	605.9
39	787.3	641	692	710.3

40	570	681.1	640.2	759.2
41	746.3	525.3	676.6	545.4
49	658.4	753.1	624.1	913.1
62	684	705.12	735.9	678.45
63	669.4	833.7	614.3	935.5
72	842.3	959	824	1081.7
88	918.8	789	989.8	882.9
314	595.8	601	569.4	631.8
326	704.4	902.2	648.2	835.4
340	678.3	670.74	651.56	729.52
341	575.6	1131.3	537.3	1395.7
345	626.2	668.06	605.7	727.86
364	509.8	458.5	580.4	490.4

Peak Torque of the arm (ft·lbs)

Subject	Peak Torque dominant arm week 0	Peak Torque dominant arm week 32	Peak Torque non-dominant arm week 0	Peak Torque non-dominant arm week 32
1	110.1	124.6	98.6	134.7
2	127.9	123	120.9	130.5
3	62.7	87.8	66.5	94.9
12	104	140.8	106.2	135.6
13	82.6	106.9	88.4	121.5
16	103.6	94.3	96.4	103.8
30	103.3	110.8	90.2	112.2
31	95.8	107.1	96.3	110.1
32	112.6	106.4	84.2	120
33	110	114.6	104.3	117.9
36	98.4	101.7	104.7	110
42	94.7	103.4	100.2	105.6
50	91	104.8	97.7	123.6
52	103.8	109.5	92.2	110.3
61	90.3	110.3	81.5	116.2
64	92.7	131	91.6	143.8
65	123.4	114.84	124.6	121.65
67	121.5	113.7	123.2	121.9
81	80	85.4	83.4	114.2
83	122.1	112.6	110.5	128.9
87	103.3	126.6	120.7	110.2
94	116.9	108.3	109.4	125.8
308	123.4	92.4	102.9	89.6
330	84.4	92.5	90.4	95.7
335	121.6	105.5	130	103.7
352	110.1	114.1	100.4	110.4
354	117.9	99.9	112.4	94.8
361	101.25	114.9	96.9	114.5
363	105.5	104	102.9	102.3
5	166.9	162.1	184.7	163.4

11	127.9	172.2	112.4	206.9
14	112.2	118.4	100.8	111.3
25	101.1	144.9	107.1	135.2
37	173.7	168.3	155.3	178.2
38	97.2	89.8	95.6	108.4
39	120.4	116.4	90.2	136
40	92.4	124.6	92	132.3
41	137.7	101.4	106.4	111.1
49	106.9	135	108.6	164.1
62	124.5	117.44	129.7	114.69
63	93.3	135	100.7	157.5
72	125.8	165.3	123.5	186.4
88	147.7	134.2	148.2	140.5
314	87.1	87.3	79.4	93.4
326	114.6	153.2	103	132.5
340	133.89	109.52	104.84	114.46
341	80.1	95.7	72.5	107.5
345	106.3	132.6	92.5	159.1
364	86.1	90.4	101.6	88.7

Total Body BMC (g/cm)

Subject	Total body BMC week 0	Total body BMC week 32
1	2298.99	2288.09
2	2090.39	2092.3
3	1747.65	1766.85
12	2047.27	2078.55
13	1891.56	1845.75
16	2033.22	2046.29
30	2315.01	2294.35
31	1894.53	1894.21
32	1858.95	1866.68
33	1920.46	1931.73
36	1943.83	1959.6
42	1985.94	2009.85
50	2125.69	2087.52
52	2053.59	2107.45
61	2321.34	2310.2
64	2136.16	2157.5
65	2310.31	2168.09
67	2587.88	2172.12
81	1640.81	1649.72
83	2279.36	2272.23
87	1845.22	1843.26
94	2654.89	2701.88
308	2649.08	2659.88
330	1835.41	1890.76

335	2022.41	2067.52
352	2593.87	2587.35
354	1914.64	1951.97
361	1920.98	1934.25
363	1867.99	1906.96
5	2287.17	2351.47
11	2594.44	2570.85
14	2350.8	2337.91
25	2169.43	2173.49
37	2201.31	2222.7
38	1741.78	1705
39	2184.09	2160.4
40	1976.61	1964.47
41	2073.82	2075.46
49	2150.6	2145.81
62	2189.41	2192.3
63	1878.32	1879.4
72	2457.63	2441.8
88	2302.53	2298.73
314	1859.69	1908.21
326	2888.84	2842.95
340	2451.71	2520.13
341	2021.66	2072.57
345	1843.64	1946.51
364	2272.16	1960.34

Total Body BMD (g/cm^2)

Subject	Total body BMD week 0	Total body BMD week 32
1	1.173	1.181
2	1.05	1.024
3	1.047	1.044
12	1.141	1.15
13	1.038	1.051
16	1.09	1.099
30	1.155	1.15
31	0.986	0.98
32	1.068	1.08
33	1.072	1.068
36	1.074	1.055
42	1.109	1.092
50	1.128	1.129
52	1.022	1.054
61	1.17	1.181
64	1.146	1.138
65	1.094	1.122
67	1.229	1.123

81	0.99	0.987
83	1.185	1.186
87	1.08	1.097
94	1.172	1.174
308	1.36	1.34
330	1.055	1.07
335	1.142	1.151
352	1.226	1.231
354	1.108	1.133
361	1.084	1.092
363	1.034	1.045
5	1.094	1.103
11	1.261	1.267
14	1.214	1.225
25	1.072	1.064
37	1.149	1.131
38	0.938	0.945
39	1.125	1.104
40	1.122	1.132
41	1.096	1.11
49	1.104	1.115
62	1.107	1.102
63	1.039	1.043
72	1.158	1.169
88	1.12	1.121
314	0.999	1.003
326	1.313	1.322
340	1.2	1.231
341	1.092	1.098
345	1.013	1.043
364	1.144	1.082

Total Ulna BMC (g/cm)

Subject	Total ulna BMC dominant arm week 0	Total ulna BMC dominant arm week 32	Total ulna BMC non-dominant arm week 0	Total ulna BMC non-dominant arm week 32
1	5.33	5.3	5.11	5.25
2	4.69	4.81	4.71	4.75
3	3.97	3.95	3.8	3.9
12	4.58	4.49	4.41	4.57
13	4.59	4.72	4.31	4.36
16	4.62	4.66	4.5	4.4
30	5.45	5.54	5.31	5.43
31	4.12	3.94	3.86	3.98
32	4.42	4.56	4.72	4.86
33	4.62	4.63	4.07	4.21
36	4.34	4.39	4.3	4.38
42	3.84	3.8	3.71	3.78

50	4.16	4.21	4.18	4.41
52	4.69	4.89	4.78	4.67
61	5.38	5.24	5.39	5.42
64	4.88	4.87	4.62	4.73
65	5.44	4.97	5.53	4.87
67	4.7	4.99	4.4	4.89
81	4.05	3.94	3.56	3.66
83	5.33	5.48	5.41	5.65
87	3.8	3.87	3.76	3.7
94	6.29	6.17	5.76	5.91
308	5.62	5.7	5.44	5.58
330	3.88	4	3.78	3.92
335	4	4.08	3.76	3.99
352	6.08	6.25	5.51	5.89
354	4.65	4.91	4.35	4.66
361	4.91	5.11	4.9	4.99
363	4.7	4.94	4.9	5.1
5	5.28	5.31	5.45	5.91
11	4.78	4.7	4.52	4.52
14	4.52	4.58	4.12	4.31
25	5.06	5.41	4.9	5.21
37	4.73	4.73	4.63	4.8
38	3.66	3.69	4.15	4.02
39	5.1	5.46	4.92	4.83
40	4.36	4.31	4.32	4.42
41	5.33	5.58	4.87	5
49	5.39	5.29	5.1	5.3
62	5.13	5.21	5.01	4.89
63	3.66	3.76	3.88	3.94
72	5.66	5.62	5.29	5.42
88	5.31	5.4	5.32	5.3
314	4.35	4.53	4.64	4.87
326	7.39	7.54	6.91	7.01
340	5.54	5.63	5.1	5.26
341	4.5	4.65	4.52	4.62
345	4.85	4.94	4.38	4.66
364	5.17	5.3	4.98	5.19

Total Ulna BMD (g/cm^2)

Subject	Total ulna BMD dominant arm week 0	Total ulna BMD dominant arm week 32	Total ulna BMD non-dominant arm week 0	Total ulna BMD non-dominant arm week 32
1	0.552	0.547	0.528	0.547
2	0.482	0.481	0.501	0.492
3	0.482	0.486	0.464	0.466
12	0.573	0.584	0.573	0.588
13	0.544	0.566	0.525	0.529
16	0.543	0.542	0.531	0.535

30	0.588	0.596	0.59	0.581
31	0.463	0.468	0.455	0.478
32	0.549	0.563	0.547	0.577
33	0.495	0.514	0.476	0.495
36	0.498	0.502	0.516	0.507
42	0.46	0.462	0.44	0.471
50	0.498	0.5	0.501	0.505
52	0.513	0.519	0.511	0.518
61	0.58	0.574	0.575	0.579
64	0.556	0.553	0.546	0.563
65	0.515	0.545	0.52	0.539
67	0.549	0.546	0.542	0.539
81	0.542	0.54	0.494	0.524
83	0.554	0.557	0.547	0.568
87	0.521	0.526	0.524	0.523
94	0.567	0.559	0.528	0.543
308	0.579	0.581	0.578	0.606
330	0.517	0.522	0.484	0.506
335	0.518	0.524	0.503	0.531
352	0.605	0.614	0.564	0.583
354	0.579	0.616	0.552	0.591
361	0.538	0.546	0.535	0.558
363	0.537	0.533	0.535	0.544
5	0.54	0.542	0.526	0.55
11	0.554	0.555	0.523	0.53
14	0.578	0.613	0.554	0.588
25	0.544	0.551	0.547	0.55
37	0.502	0.5	0.512	0.508
38	0.484	0.482	0.551	0.547
39	0.579	0.636	0.564	0.575
40	0.487	0.479	0.481	0.483
41	0.549	0.567	0.534	0.542
49	0.572	0.582	0.56	0.592
62	0.516	0.507	0.493	0.492
63	0.471	0.48	0.484	0.485
72	0.58	0.568	0.542	0.542
88	0.544	0.539	0.524	0.528
314	0.509	0.522	0.502	0.523
326	0.674	0.704	0.626	0.647
340	0.497	0.513	0.493	0.506

Ulnar Bending Stiffness (N·m²)

Subject	EI dominant arm week 0	EI dominant arm week 32	EI non-dominant arm week 0	EI non-dominant arm week 32
1	24.1	25.9	25.1	34.3
2	23.6	20.9	12.5	22.3
3	8.7	6.6	17.4	10.1

12	16.9	22	15.4	21.4
13	21.5	23.8	10	24.8
16	27	35.2	52.8	42.1
30	21.6	33.2	27.1	28.6
31	24.2	20.3	34.1	22.5
32	12.8	13.5	18.8	16.8
33	40.4	22.8	14	14.5
36	20.7	21.1	14.1	16.5
42	10.9	14.6	15.9	13.4
50	32.7	25	20.2	21.4
52	9.8	13.7	30.1	24.2
61	13.5	18.4	23.8	32
64	25.4	28	23.5	16
65	24.8	24.2	29.4	23.2
67	11.7	18	13	15.5
81	2.7	12.7	7.3	17.7
83	27.1	18.2	16.6	29.6
87	24.2	26.3	44.5	26.5
94	22.5	10	14.2	39.7
308	23.8	24.8	20.2	11.1
330	21.5	21.1	11.6	12.1
335	10.7	18.8	21.7	14.2
352	18.3	26.4	17.2	34.1
354	21.7	23.1	26.3	34.5
361	19.8	23.3	12.7	36.3
363	14.8	29.7	14.2	23.8
5	18.8	22.6	17.6	32.2
11	16.9	17.6	10.1	18
14	17.7	18.3	18.2	12.4
25	21.9	21.1	20	24
37	20.7	15.2	10.4	20.1
38	20.8	22.5	19.4	18.8
39	18.5	16	3.3	25.6
40	16.3	11.7	11.2	45.2
41	20.9	14.8	5.3	8
49	26.2	26.3	25.1	19.9
62	27.8	44.7	30.8	39.5
63	19.9	12.8	22.5	13
72	18.7	21.1	17.6	21
88	10.8	25.7	10.4	23.4
314	21.4	11.6	13.5	19
326	29.1	39.8	35.3	33.9
340	24.7	24.4	10.6	28.8
341	24.6	35	14.7	21
345	2.7	3.1	3.9	2.4
364	10	22.5	20	18.3

Vita

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EDUCATION

M.S., Clinical Exercise Physiology, May 2004
Virginia Polytechnic Institute and State University, Blacksburg, VA

B.S., Science of Food Nutrition and Exercise, May 2001
Virginia Polytechnic Institute and State University, Blacksburg, VA

Research Experience:

2000-2003: Laboratory Assistant; Laboratory for Health and Exercise Science, Virginia Polytechnic Institute and State University, Blacksburg, VA.
2003-present: Laboratory Assistant; Muscle Physiology and Biochemistry Laboratory, Blacksburg, VA.

Academic/Teaching Experience:

2002-present: Teaching Assistant; Virginia Polytechnic Institute and State University, Blacksburg, VA.

RESEARCH INTERESTS

-Resistance training and bending stiffness in the human ulna.
-Fracture risk assessment by mechanical testing (MRTA).
-Pathogenic mechanism associated with Duchenne's Muscular Dystrophy
-Gene regulation in the *mdx* mouse model compared to wild type controls

ABSTRACTS

Mott S, Williams B, Wootten DF, Ramp WK, Kelso T, Herbert WG, FACSM. Do arm muscular strength variables influence ulnar bending stiffness? Med Sci Sport Exer 2002;34(5):S147.

B.O. Williams, K.D. Zeff, S.M. Nickols-Richardson, L.E. Miller, L.M. Pierson, D.F. Wootten, W.K. Ramp, and W.G. Herbert, FACSM. Bone-Free

lean mass and strength adaptations with eccentric vs. concentric isokinetic strength training in young women. Abstract submitted to: American College of Sports Medicine (ACSM) International Conference, St. Louis, Missouri, 2004.