The Winthrop McNair Research Bulletin

Volume 2 *The Winthrop McNair Research Bulletin Volume 2*

Article 7

2016

Comparing the tensile and compressive Young's moduli of cortical bone

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ABSTRACT

Various methods have been used in the testing of the mechanical properties of cortical bone, specifically the young's modulus. However, in the case of the material's Young's modulus in compression and tension, there is a significant disagreement among the published findings that may be a result of experimental artifacts. This study attempts to solve the scientific question of whether cortical bone is stiffer in compression or tension and if so to understand why that is. Using small samples taken from the distal portion of the neck of a femur belonging to a young white tailed deer, both the Young's moduli in tension and compression were calculated for each sample. The sample was placed under loads that wouldn't result in plastic deformation [max load of 140 Newtons) thus allowing the ability to use the same sample for both tension and compression without compromising its mechanical properties. Results show that elastic modulus is greatest while in tension, specifically in the load bearing cranial orientation.

INTRODUCTION

Due to the lack of standardization or consistent testing methods, previous studies on cortical bone have no defined conclusion as to whether it is stronger in tension or compression. Some sources state that cortical bone in compression is stronger, some suggest that tension is greater, while others state that there is no significant difference at all. It is difficult to gain validation or compare the works of various authors because each experiment was carried out using different methodical procedures. With so many uncontrolled parameters in play, this makes it practically impossible to compare at eyes view. In addition to this, calculations of the Young's modulus in cortical bone can vary across species, across different bones within that species and even across different regions of the same bone. Different samples have the potential to have completely different mechanical properties. This is why the data collected from samples measured in tension can not be easily compared to those measured in compression, even if the same methods and procedures were performed. Our study attempts to neutralize or minimize these possible errors, resulting in a more controlled experiment by using the same sample tested in both tension and compression. Taking this approach guarantees that the material composition is exactly the same for each individual test subject measured in tension and compression.

Bone Composition

Bone is made up of a combination of organic and inorganic elements. The mineral component is composed of insoluble salt crystals called hydroxyapatite. Hydroxyapatite, which makes up roughly 60% of the bone, contains large volumes of calcium and phosphate minerals along with traces of other minerals such as magnesium, sodium, and bicarbonate. This, together with organic collagen fibers, gives bone its strength and rigidly. The remaining 20% is composed of water, like all living tissues, and provides some flexibility to the rigid backbone.

Mechanical Properties

Thanks to this unique composition, bone exhibits viscoelastic behavior, meaning that unlike completely stiff and elastic materials, bone dissipates energy in the form of heat as it is loaded in an attempt to regain equilibrium. Thus the strain rate is dependent upon time, allowing the bone to gradually return to its original shape after a load has been applied. In addition to this viscoelasticity several other properties affects the bone's mechanical properties. Every bone is unique with the ability to change its composition based on its environmental feedback. In accordance with renowned German anatomist Julius Wolff, Wolff's Law states that bone will adjust its structure to better accommodate mechanical loads placed upon it. Bones that are under high stress may contain more inorganic minerals while those that are under lower levels of stress may have a composition that contains more collagen and water. Living bone (In vitro) is constantly changing and works in an endless feedback loop to continuously optimize its mechanical behavior. Mineral content contributes a lot to a bone's strength, but can also be affected by a number of factors including the amount of water and the presence or absence of bone diseases such as osteoporosis which degenerates and changes the architecture of the bone.

Application

Conclusive evidence of this scientific question has substantial use in applications such as bone grafting and prosthetics. Bone disease such as osteoporosis is a growing epidemic worldwide. An estimated 1.5 million individuals suffer a fracture due to bone disease each year. With a constantly aging population, this will be a problem for years to come. Roughly 50 million individuals over age 50 have bone degeneration of the hip and are at risk of complications later in life. It is projected that by 2020, one in two Americans over age 50 will have or be at risk of suffering from some sort of bone disease. Medical advancements such as hip implants and bone grafts have allowed individuals with bone disease to have a better quality of life. These materials that are put in the place of bone attempts to replicate its function; however they can not do so without properly knowing the properties of the bone itself. This is imperative especially for implants because they can't change its mechanical properties like bone can. It also affects the surrounding tissues and could potentially result in even more serious problems if the implant fractures or if it is too stiff. This would cause the bone to resorb due to stress shielding and thus further accelerate osteoporosis.

Young's Modulus

Young's modulus, also known as elastic modulus, is used to measure the ability of a solid material to endure changes in length while under lengthwise tension or compression. This helps to provide a numerical value to describe a material's mechanical properties. It is expressed by the following equation: Young's Modulus = $(FL_0)/A(L_n - L_0)$ and defines the relationship between stress (force per unit area) and strain (proportional deformation). A force F in the form of tension or compression is applied to a specimen at each end with a cross-sectional area A. This causes the specimen to change its original length L_o to some new length L_n. The stress is defined as the quotient of the tensile force divided by the cross-sectional area, or F/A and the strain or relative deformation is the change in length, $L_n - L_0$, divided by the original length, or $(Ln - L_0)/L_0$. A solid body deforms when excessive loads are applied to it. The material displays elastic behavior if the body returns to its original shape after the load is removed, thus signifying that no deformation has occurred. Bone is known to start displaying plastic deformation under loads exceeding 150N. For experimental purposes, the specimen will undergo max loads of 140N in order to avoid plastic deformation, allowing the ability to reuse the specimen without compromising its mechanical properties.

MATERIALS AND METHODS

The right femoral bone was harvested from a white tailed deer that died to causes unrelated to failure of the musculoskeletal system. Using a handsaw, a 40 mm portion of its diaphysis was taken from the distal portion of this bone and the interior bone marrow was removed. Following this the small shaft was divided into four equivalent regions, each belonging to each orientation of the bone; Cranial (Front), Caudal (Rear), Medial (Side closest to the body) and Lateral (Side farthest from the body). Doing so helps to see if there is any difference in the mechanical properties of the bone based on the loading direction, which is basically checking for proof of Wolff's Law. Wolff's Law states that that bone grows and remodels in response to the forces that are placed upon it. Specific directions experience different stresses that the body picks up and remodels the bone to become most efficient at supporting the load. From each region four 2x2x40mm samples were cut using a Allied TechCut4 Bone saw with a 4x.012x.05in diamond metal wafering blade. To do so, the bone had to be binded to the saw using an inorganic Jet acrylic that was made in the lab from a liquid powder resin mix.

To better simulate wet bone, each sample was allowed to soak in water 24hrs prior to testing. Individually the samples were taken and loaded in tension on an Instron 5942 machine with only 4mm of the sample exposed, which was the test site. A load of 0-140N was placed on the sample at a strain rate of .5. These values apply loads that are similar to real-life expectations and at a desirable rate to ensure that the bone won't fracture. This all occurs within the elastic region of bone, meaning that under these parameters no deformation of the bone will occur. From this test, the Young's Modulus was gathered and stored. The sample was unloaded, reloaded and tested a second time to confirm the accuracy of our data where similar results minimized the chance of slippage.

Immediately following this, the sample was cut into a 2x2x4mm beam, the same exposed region that was tested in tension. The Instron machine is restructured to test in compression and the now smaller beam is loaded. To keep the beam stabilized, a pea size amount of Filtek Z250 resin based Dental Restorative Material was added to the load sites (top and bottom of the beam). This composite was polymerized, exposing it to UV light. Each load site was treated for 30 sec. Using the same parameters, 140N max load, .5 strain rate, the Instron machine can now test the same sample in compression. This setup can be seen in figure 1. Once again the Young's Modulus was calculated twice, once for each time the procedure was ran. After all of the samples were measured in both tension and compression, the results could be compared.

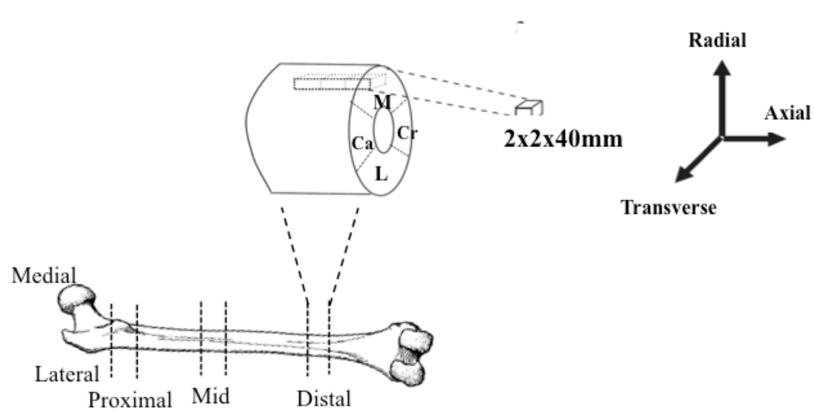

Figure 1: Figure showing the 40 mm section of the femur the test samples were retrieved from as well as how the caudal, cranial, medial and lateral quadrants were established.

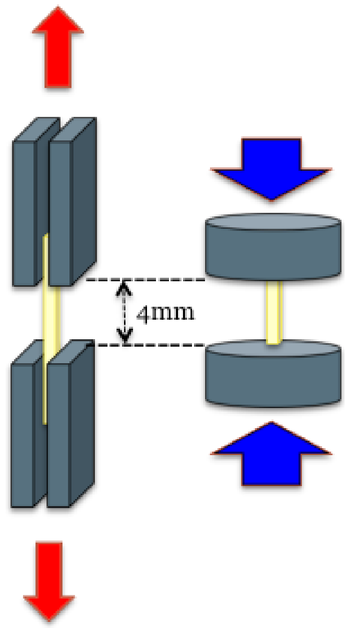

Figure 2: Diagram showing the setup of the Instron machine. The specimen was first placed inside the instron which pulled the sample, applying tensile forces. As the material is being pulled, its elongation could be automatically observed and documented by the Instron machine. Over time this will output a resulting curve or tensile profile showing how the materials react to the forces being applied. The sample was cut and repeated in compression.

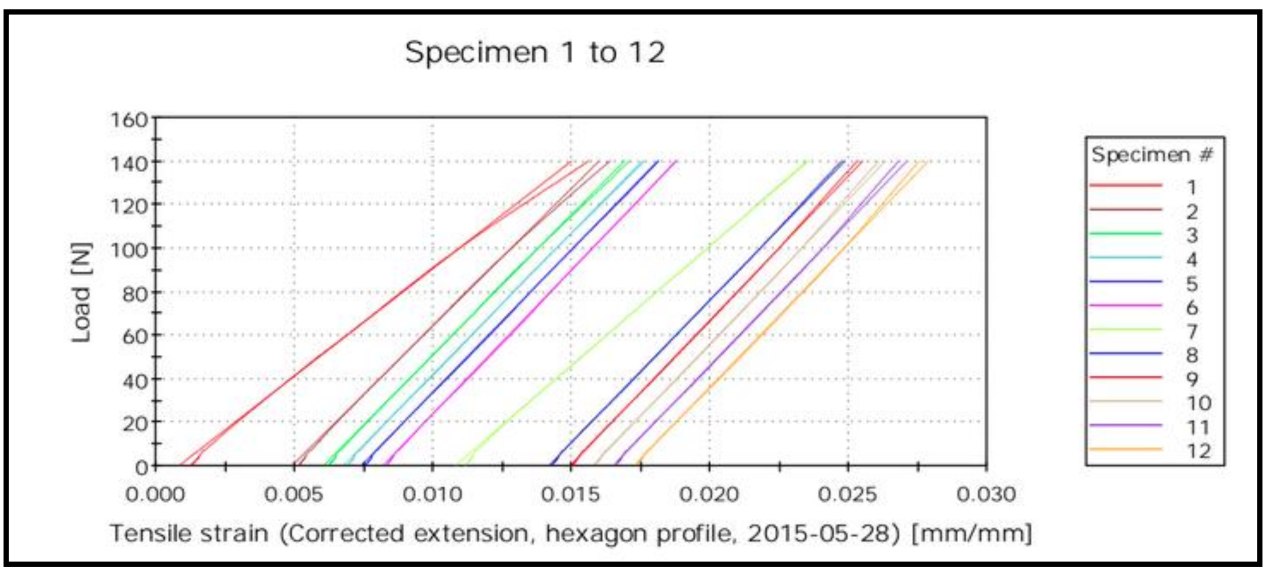

Figure 3: Original data showing the continuous elastic behavior of the specimen tension under cyclic loading.

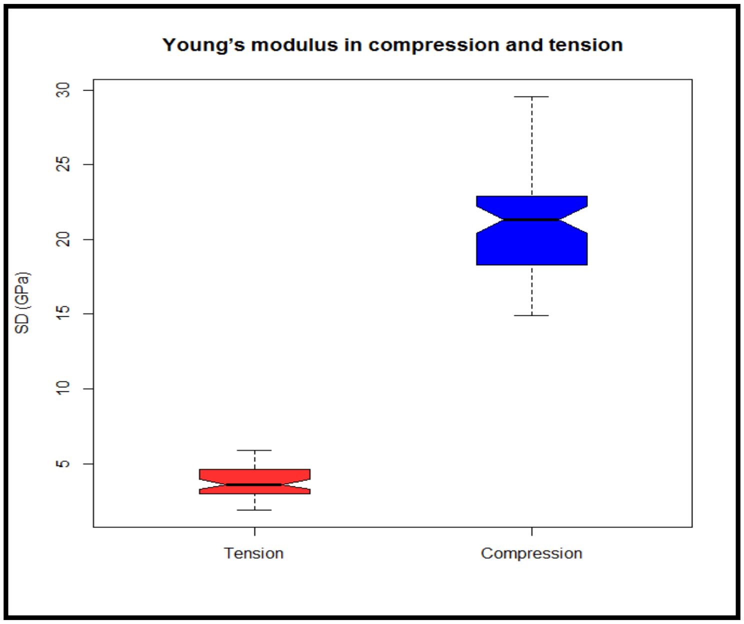

Figure 4: The box and whisker plot shows that the stiffness of bone in Compression is significantly greater than that in tension.

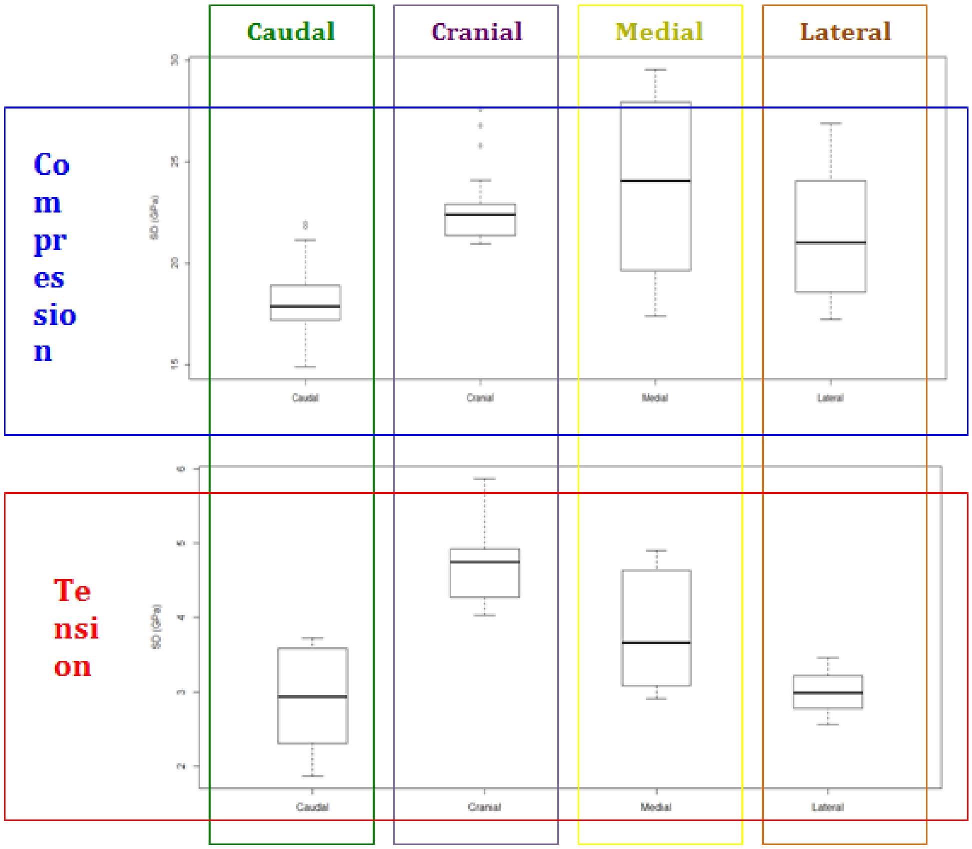

Figure 5: Diagram comparing the Young's modulus of cortical bone in tension and compression by region (Caudal, Cranial, Medial and Lateral).

DISCUSSION AND CONCLUSION

As seen in figure 4, the Young's modulus of cortical bone was significantly higher in compression than in tension. Comparing the quadrants, shown in figure 5, the stiffness of cortical bone in compression for the caudal region is much lower than the other three orientations, which are relatively similar. However, when compared in tension, the cranial region was the stiffest while the others were similar. Extrapolating the data, the cranial region showed the highest stiffness, followed by the medial and lateral region (tie) and lastly the caudal region. While cortical bone influenced heavily whether it is in tension or compression, the orientation that the bone comes from has a significant affect as well, specifically the cranial and caudal regions. These findings would have to be repeated for accuracy, as there are a number of factors that could have influenced the results. This includes but is not limited to, not having enough samples to establish a normally distributed curve, the inability to measure compression before tension, or having inadequate hydration of the samples before loading. As stated previously, viscoelastic materials release energy as heat, so when force was applied in tension, it may have dehydrated the sample, making it more stiff and brittle. The samples were cut and tested after only 30 minutes of rehydration (submerged in water), which may not have been long enough. While further studies would have to be done, the data suggest that the elastic modulus is greatest while in tension, specifically in the load bearing cranial orientation.

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