Published in final edited form as:

J Orthop Res. 2014 December ; 32(12): 1580–1588. doi:10.1002/jor.22720.

Experimental and Finite Element Analysis of Dynamic Loading of the Mouse Forearm

Ganesh Thiagarajan¹, Yunkai Lu¹, Mark Dallas², and Mark L. Johnson²

¹Department of Civil and Mechanical Engineering, University of Missouri-Kansas City, 350L Flarsheim Hall, 5100 Rockhill Road, Kansas City, Missouri 64110

²Department of Oral and Craniofacial Sciences, School of Dentistry, University of Missouri-Kansas City, Room 3143, 650 E 25th Street, Kansas City, Missouri 64108

Abstract

Bone formation is reported to initiate in osteocytes by mechanotransduction due to dynamic loading of bone. The first step towards this is to characterize the dynamic strain fields in the overall bone. Here, the previously developed mouse forearm ulna-radius model, subjected to static loading, has been further enhanced by incorporating a loading cap and applying a cyclic dynamic load to more closely approximate experimental biological conditions. This study also incorporates data obtained from strain gauging both the ulna and radius simultaneously. Based on separate experiments, the elastic modulus of the ulna and radius were determined to be 13.8 and 9.9 GPa, respectively. Another novel aspect of the numerical model is the inclusion of the interosseous membrane in the FE model with membrane stiffness ranging from 5–15 N/mm that have been found to give strain values closer to that from the experiments. Interestingly, the inclusion of the interosseous membrane helped to equalize the peak strain magnitudes in the ulna and radius (~1800 at 2 N load and ~3200 at 3.5 N), which was also observed experimentally. This model represents a significant advance towards being able to simulate through FE analysis the strain fields generated in vivo upon mechanical loading of the mouse forearm.

Keywords

mouse forearm; finite element analysis; experiments; dynamic load; interosseous membrane

Bone adjusts its mass and architecture in response to changes in the load it experiences. Over the past several years, many studies have been devoted to understanding the mechanism(s) by which bone senses and responds to mechanical signals. Ozcivici et al.¹ conducted a comprehensive review that explains how bones transduce mechanical signals into anabolic or catabolic responses through the process of mechano-sensing. Generally, the osteocyte is believed to play a key role in this process.²

The adaptive response of cortical bone to loading is largely influenced by the magnitude of load or mechanical strain with new bone formation occurring^{3,4} once the strain exceeds a

threshold value. However, only dynamic excitation can initiate an anabolic response in bone and static excitation can even suppress bone formation,^{5,6} which indicates that cortical bone adaptation is not dependent on strain magnitude alone. Osteocyte responses to load have also been hypothesized to be driven by the shear force generated by the gradient produced from the interstitial fluid flow in canaliculi.⁷ This hypothesis supports the concept that bone adaptation occurs only under dynamic loads since static loads will not initiate fluid movement in the cell.

Other parameters which have also been investigated that influence the anabolic response to loading include loading frequency,^{8,9} strain rate,^{10,11} strain gradient,¹² loading patterns,^{13,14} and loading duration.^{15,16} Loading frequencies ranging from 0.5 Hz up to 30 Hz have been compared in these studies. Various researchers have also investigated the effects of fatigue loading on bone fracture.^{17,18}

The ulna compressive loading model has been widely used to study bone formation in response to mechanical loading.^{6,10,19–21} Few attempts have been made to build numerical models to examine the effect of loading frequency on the bone response. In vivo mouse compression loading experiments are typically conducted by applying a cyclic load that produces a particular maximum bone surface strain in the ulna or tibia. In these experiments, the periosteal strain in the longitudinal direction on the loaded bone is measured using a strain gauge attached to the bone surface. In order to understand the mechanisms by which in vivo loading generates an osteogenic response, FE models have been constructed to assess general strain distributions within the bone tissue that result from the applied external mechanical loading. These models are routinely calibrated using the strain data acquired from strain gauges placed on the ulna and provide an estimate of the strain distribution within the ulna in vivo. FE models of the mouse tibia^{22–24} and rat ulna^{17,25,26} have been described by various researchers, but dynamic models of the mouse forearm have not been reported in the literature thus far.

The overall goal of our research is to study the mechanical strain environment at the osteocyte cellular level under dynamic in vivo loading. Changes in osteocyte signaling pathways and gene expression have been shown to occur in response to in vivo and in vitro loading (see Review by Bonewald and Johnson²⁷). Hence, we believe that the dynamic strain history that the osteocyte experiences influences the pathway activation. This work is a first step towards the characterization of the dynamic strain history numerically using a macro bone model. In order to achieve this goal, a dynamic model of the ulna and radius must first be developed. The objective of this study is to examine experimentally and numerically the full mouse forearm model when subjected to the cyclic dynamic loading in order to characterize the dynamic strain history in the bone. The experimental objectives were achieved through cyclic compression loading of the mouse forearm and performing three point bending experiments on the ulna and radius separately in order to determine their individual macro elastic modulus. This study also employed two different load magnitudes and loading frequencies to assess the effect of preload on strain. Both the ulna and the radius were strain gauged and to the best of the author's knowledge, this is the first reported attempt to place strain gauges on both bones. This work introduces three new concepts from a modeling perspective: (a) the inclusion of the interosseous membrane; (b) the addition of a

cap with boundary conditions that approximate the loading conditions in the mouse forearm; and (c) the simulation of cyclic dynamic loading. The results obtained here help provide a theoretical prediction as well as comparisons for future experimental work in studying bone adaptation to mechanical loading.

METHODS

Experimental Measurement of Strains and Modulus

Various studies have reported utilizing strain gauges mounted on both the ulna and radius of sheep²⁸ and dogs.²⁹ Other works have utilized one strain gauge such as rat ulna²⁵ and mouse ulna.^{30–32} As shown in Figure 1, strain gauges were attached to both the ulna and radius of the wild-type six-month-old female TOPGAL mouse forearm (C57Bl/6 x CD1 mixed genetic background). The strain gauging procedures as outlined in Robling et al.³³ were followed in this study. Experimental data was obtained from ex vivo loading using the Bose ElectroForce 3220 system (Bose Corp., ElectroForce Systems Group, Eden Prairie, MN). Strains were measured both on the ulna and radius using trimmed single element strain gauges (EA-06-015DJ-120/LE, Vishay Micro-Measurements, Raleigh, NC) applied to the medial-surface of a single female mouse ($n = 1$) forearm at age six months. The contact area for the strain gauges were approximately 1 mm $\times$ 2 mm.

The following experimental protocol was adopted to account for the effects of load magnitude, loading frequency, and preload level. First, both the ulna and radius were imaged using a microCT scanner (Scanco μ CT, Scanco Medical, Basserdorf, Switzerland) and then strain gauges applied. Ex vivo compression tests were performed with two load levels (2 N and 3.5 N) and two frequencies (0.2 Hz and 2 Hz) at two different preload levels (0.3 N and 0.8 N) for a total of six compression tests on the bone. Subsequently, the ulna and radius were separated and subjected to three point bending tests in order to obtain the individual macro elastic modulus.

Finite Element Modeling

FE models of the wild-type mouse forearm were developed from μ CT images with a 65 μ m in-plane resolution and 12 μ m axial slice spacing. DICOM images were imported into Slicer3D (<http://www.slicer.org>)³⁴ to segment and create models of the ulna and radius. These models were then imported into GeoMagic Studio 9 (3D Systems – Geomagic Solutions, Morrisville, NC) for smoothing, patching, curve fitting, and surface mapping before the final CAD models were generated. The loading cap was imported as an IGES file and aligned with the bone to create a combined geometry. Finally, volume meshing was performed using an automatic mesh generation process in Abaqus/CAE (Dassault Systèmes, Vélizy, France) using tetrahedral elements. Finite element analysis (FEA) was performed in LS-DYNA v71 (Livermore Software Technology Corp., Livermore, CA) using element formulation number 10 (4-noded tetrahedron with 1 integration point). The mesh of the mouse forearm consisted of 192,674 nodes and 124,145 tetrahedral elements while the mesh of the cap consisted of 2322 nodes and 1112 eight-node hexahedral elements. The same mesh seed size was chosen in all directions. Other studies have shown tetrahedral elements

to be fairly effective in bone FE simulations.³⁵ Mesh sensitivity studies conducted previously²⁴ have also shown convergence for the mesh size selected for this study.

The distal and proximal end of the ulna and the radius, connecting the coronoid process of the ulna to the head of the radius and the styloid process at the distal end, were connected at a few nodes using linear one dimensional spring elements. A spring constant of 1400 N/mm was selected initially³⁶ but values of 1000 N/mm and 1750 N/mm were also used to assess the effect of varying the spring constant. The average peak compressive strain values, for an applied force of 2 N, in the medial region of the mid shaft of the ulna did not vary much with the change in the spring constant value (10,250–10,260 $\mu\epsilon$). The interosseous membrane was modeled as a criss cross pattern of spring elements and a parametric study was conducted with stiffness values ranging from 2.5 N/mm to 15 N/mm.³⁷ The cortical bone mineral density was 1168 kg/m³ for the ulna and 1172 kg/m³ for the radius. The Poisson's ratio was 0.3 and 0.3 for ulna and radius, respectively.

The dynamic model shown in Figure 2 is an extension of the authors' previous static model²⁴ that incorporates a loading cap to more closely simulate experimental conditions. Dynamic cyclic loads of magnitudes 2 N and 3.5 N were applied at two different frequencies (0.2 Hz and 2 Hz) and two different preloads (0.3 N and 0.8 N) at the distal end through the loading cap while the proximal end of the ulna was fixed at various nodal locations. Surface to surface contacts were defined for the cap/ulna and cap/radius interfaces, respectively. Explicit dynamic analysis in LS-DYNA was used to run the simulations, for two cycles with the time-step controlled internally by the program based on the element size. Simulations of two cycles of loading at 0.2 Hz frequency were completed within a reasonable time, approximately 3 to 4 h on a modern desktop machine, using mass scaling and satisfying the Courant time-step criterion. The Courant time step criterion was used to develop an internal time step to be analyzed. This time step ensures that the time step chosen for analysis was small enough for the explicit dynamic solution to converge to a corresponding implicit solution. Mass scaling is a way to artificially increase the density of the material in order to keep the Courant time step large enough to complete the explicit analysis in a few hours. Not using mass scaling could result in each FE analysis taking days or more to complete one analysis. Energy due to this artificial increase in mass is monitored in order to ensure that inertial forces due to the mass increase do not affect the final results. In this analysis we did not encounter this concern (Fig. 3).

Material Properties of Bone

Three point bending experiments were performed on the ulna and radius separately and two experiments were performed on each bone. The first employed only a small load while the second involved loading to failure. The CAD models of the ulna and radius were then imported into SolidWorks (Dassault Systèmes SolidWorks Corp., Vélizy, France) in order to measure the cross-sectional properties and calculate the moments of inertia (MOI) at seven different sections, as seen in Figure 4, along the length of the bone between the two supports. The elastic modulus was then determined using the recorded displacement value at the load point and the elastic deflection equation (1), where P is the magnitude of the force,

L is the length between the two fixtures, I is the moment of inertia (second moment of area), and Δ is the deflection.

$$E = \frac{PL^3}{48I\Delta} \quad (1)$$

A FE analysis of the ulna and radius was performed separately to mimic the three point bending experiments. In this analysis the input values for the elastic modulus were varied parametrically until the deflections from the experiments and simulations matched. Since experiments were carried out using the three point bending test for the ulna and radius loaded to failure separately, FE simulations were run with each of the bones loaded to their respective experimental failure load. FE models were then analyzed at this failure load to determine the tensile strain magnitudes in both the ulna and radius.

RESULTS

Experimental Strain Data

Experiments were conducted at two applied loads (2 N and 3.5 N), two frequencies (0.2 Hz and 2 Hz), and two preloads (0.3 N and 0.8 N). Figure 4 displays the strain gauge readings from the 2 N load tests for a preload level of 0.3 N and 0.8 N while Figure 5 shows the strain readings for the 3.5 N load and preload of 0.3 N. As shown in Table 1, variations in the frequency did not generate significantly different strains in the ulna and radius. The curves shown in Figures 4 and 5 along with the data in Table 1 also suggest that the magnitude of strains are approximately the same in both the ulna and radius.

Studies were next conducted to examine the effect of preload levels on the strain gauge readings (Table 1), and it was determined that the magnitude of peak strain decreased as the preload increased. The range of the force range is reduced by 0.5 N when the pre-load is increased from 0.3 to 0.8 N. For a 2 N total force, the force range is 1.7 N for 0.3 preload, and 1.2 N for 0.8 N preload. Hence, a 30% reduction in strains can be expected for the 0.8 N preload. The decrease in the strain magnitude ranged from 31% to 40% for the 2 N load and 24% to 40% for the 3.5 N load. The peak strains at the lower frequency were also lower or similar to those at the higher frequency.

Experimental Estimation of Bone Modulus and FE Verification

The μ CT scans of the forearm were used to build a CAD model for the three point bending tests. The MOI was calculated from seven cross sections located along the central region of both ulna/radius model. The average MOI (mean $\pm$ SD) was $0.01213 \pm 0.00074 \text{ mm}^4$ for the ulna and $0.01575 \pm 0.00432 \text{ mm}^4$ for the radius (Table 2) for the lateral-medial axis. Figure 6 shows the variation in MOI along the bone proximal-distal direction. In Figure 6, the x -axis represents the medial-lateral direction. Based on these plots, the radius MOI increases nearly two-fold from the proximal to the distal end, and the radius has a higher average MOI compared to that of the ulna (Fig. 7).

FE models of the ulna (16,621 nodes and 54,253 elements) and radius (19,832 nodes and 89,628 elements) were created and analyzed using a static analysis for three point bending conditions. Table 3 shows the elastic moduli for the ulna and radius that were estimated from experiments and equation (1). FE models of three point bending tests were created to verify these estimations by parametrically varying the elastic moduli until the deflection at the mid-bone segment matched the experimental data. Table 3 also shows the elastic modulus determined from FE simulations with the initial load as well as the experimental and FE results for the displacements. Centrally located elements in the FE model were selected to compute the average strain and subsequently estimate the strains in the ulna and radius at the failure load. By performing a linear FE analysis (Table 3), the peak compressive strain predicted at the experimental failure was over two times higher in the radius compared to the ulna. Furthermore, the tensile failure strain was about 1.6 times the corresponding ulna strain value. These predictions did not assume any stress or strain based yielding or failure criterion and assumed a fully elastic behavior up to failure.

Comparison of the Strain Values Between FE and Experiments

Due to difficulties in evaluating the stiffness of the springs representing the interosseous membrane, the spring constant was varied parametrically and the forces in the ulna and radius were determined from FE simulations. From the FE model pre and postprocessing unit LS-PREPOST the resultant force in any part can be determined by choosing a section location and the part and calculating the sum of all forces in the nodes in that section. For our analysis, the ulna and radius were chosen as the two sections and the resultant force in the two bones were determined. The resultant force calculated at the mid-shaft of the forearm is plotted in Figure 8 for two cycles under different sinusoidal loads. The resultant force matched the applied load relatively well at the lower stiffness of 2.5 N/mm, suggesting that the ulna and radius carried most of the force. However, the forearm experienced a dramatic drop in the load at stiffness values of 5 N/mm or greater, suggesting that the interosseous membrane also carried a percentage of the axial load applied to the bone.

The FE strain predictions were compared with strain gauge readings to assess the current dynamic model. The average strain was determined from elements that matched the location of strain gauges attached to the forearm. Figure 8 shows the magnitudes of peak average strains in each bone for different stiffness constants. In both the ulna and radius of the dynamic FE model, the peak values of the average strain plots are much higher (140%–240%) compared to the experimental counterparts depending on the stiffness value used in the simulations. Table 4 provides a summary of the peak values for the average z-strain curves of selected elements. For the 2 N case, the strains in the ulna and radius are approximately the same at each stiffness. For the 3.5 N case, a similar observation can be made especially as the stiffness increases beyond 5 N/mm. As the membrane stiffness increases to 15 N/mm, the strains tend towards the experimental values.

Figure 9 shows the contour plots of the compressive strains in the medial surface of the ulna and the radius. These results are from a simulation of 2 N peak load with a preload of 0.3 N and the interosseous membrane stiffness value of 10 N/mm. The figure shows the contour plots at times of 0.5, 1.0, 1.5, 2.0, and 2.5 s which represents one quarter cycle. The

progression of the compression region of the strain in the two bones can clearly be seen in Figure 9. Figure 10 displays a comparison of the compressive strain contour on the medial surface of the ulna and the radius for various values of the interosseous membrane stiffness. It can be seen that as the stiffness values increases the strain contour plots on the bone become less intense and the regions of high compressive strain values decreases in size.

DISCUSSION

This paper describes our experimental and numerical studies of the mouse forearm under various biologically relevant dynamic loading conditions. The results presented here are based on a single sample size. The FE model incorporates new features such as the interosseous membrane and a distal cap to distribute the load to the radius and ulna.

Loading Frequency Effect

Based on results in Table 1 and Figures 5 and 6, the variation in strain for the 2 N load, the maximum change in strain in the ulna and radius was $130 \mu\epsilon$ which is approximately 5% of the peak strain value. The results were somewhat inconsistent for the higher 3.5 N load.

Preload Effect

As the preload increased, the peak experimental strains decreased for two reasons (Table 1). Firstly, the increasing preload created a greater rigidity at the caps and reduced bone rotation at the ends, which tend to behave like a fixed joint. This higher rigidity subsequently led to lower peak strains in the member. Secondly, the percentage of load applied as preload during the 3.5 N case was lower compared to the 2 N case. Hence, the rigidity at the ends in this case was not as pronounced. The range of load applied decreases as the preload increases. Hence, it is expected that the experimental strains would decrease which was observed to be in the range of 24%–40%.

Peak Ulna/Radius Strain Ratio

The ratio of the peak average strain between the radius and ulna (Table 4) ranged between 0.97 and 1.02 for both loading frequencies under 2 N. Under 3.5 N, the ratio varied from 0.78 to 0.98 at 2 Hz and from 0.89 to 1.01 at 0.2 Hz. These ratios match their counterparts from the test data, but the magnitudes of predicted strains were higher than the strain gauge data at the lowest interosseous membrane stiffness. The mismatch between FE and experimental strain magnitudes was believed to occur because the strain values were location sensitive. The strain gauges could have been attached to a region subjected to a lower strain field. A magnitude shift in the reading for the 2 Hz loading case is observed in Figure 5, which possibly resulted in the observed peak strain difference. During actual tests, other factors could have contributed to the stiffness of the specimen. These include the adhesive and strain gauges themselves as well as the distal and proximal end connections at the paw and olecranon. Consequently, the simulation results provided an estimate of strains in the bone. The maximal compressive strain occurred approximately 3.5 mm and 3.4 mm from the distal end of the ulna and radius, respectively. However, as the interosseous membrane stiffness increased from 5 to 15 N/mm, the strain magnitudes in the two bones were not only equal but also closer to the experimental values. The dynamic modeling

approach presented in this paper included inertial effects but not viscous effects and did not explain rate-dependent effects documented in experimental studies of in vivo loading.

Interosseous Membrane Effect

Based on dynamic FE simulations, the interosseous membrane significantly influences the load distribution between the ulna and radius, and consequently, the strain distributions in the bone. Without this membrane, the load is entirely taken by the ulna and radius, as seen by the dotted lines in Figure 8. However in Figure 8, as the membrane stiffness was increased from 2.5 N/mm to 15 N/mm, the force in the ulna and radius drops from 1.65 N to 1.25 N for the 2 N axial load case and 3.25 N to 1.85 N for the 3.5 N axial load case. As a result, both the ulna and radius are subjected to smaller loads that lead to lower strain fields (Table 4). Pfaffle et al.³⁸ experimentally determined the stiffness of the human interosseous membrane to be 13.1 N/mm. Experimental and numerical results also indicated that the peak strain magnitudes in the ulna and radius were approximately the same as the interosseous membrane stiffness increased beyond 5 N/mm, and the strain magnitudes matched closely with the experiments at higher membrane stiffness values of 10–15 N/mm.

CONCLUSIONS

We have developed a dynamic FE model to predict local strain and stress values experienced by the ulna and radius subjected to various loading conditions. Interestingly, the interosseous membrane was found to be a significant factor in determining the peak strain magnitudes and strain distributions in the ulna and radius. Our successful attempt to attach strain gauges to both the ulna and radius has provided a much more biologically relevant determination of loads and resulting model of load/strain distributions in the radius and ulna without the need to incorporate estimates of load sharing between these bones in the forearm. This research will provide a solid foundation for our next step in creating biologically valid FE models to study the effect of local strain distribution on bone formation and the strains experienced by osteocytes that alter their biological behavior during dynamic mechanical loading.

Acknowledgments

Grant sponsor: National Institutes of Health; Grant numbers: NIA P01 AG039355, NIAMS R01 AR053949.

The project was funded by a grant from the National Institutes of Health — NIA P01 AG039355 and NIAMS R01 AR053949 (ML Johnson-PI).

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Figure 1.
A mouse forearm specimen under compression test with strain gauges attached to both ulna and radius.

Figure 2.

Geometric and Boundary Condition of the Ulna/ Radius Model with Loading Cap (right) and the close up view of the membrane (left). Loading is applied at the cap which distributes the forces to the ulna and the radius based on contact conditions. The olecranon end is fixed by applying pin conditions to a few nodes. The view on the left shows the crisscross pattern of the springs used to model the interosseous membrane.

Figure 3.

End sectional cuts shown used for the determination of second moment of area. A total of seven equally spaced cuts from the central region of the ulna and radius were taken separately and the second moment of area was determined using SolidWorks.

Figure 4.

Strain gauge readings under 2 N load (0.8 N preload). Strain gauge readings from both the ulna and the radius are shown for loading frequencies of 0.2 Hz and 2 Hz.

Figure 5. Strain gauge readings under 3.5 N load (0.8 N preload). Strain gauge readings from both the ulna and the radius are shown for loading frequencies of 0.2 Hz and 2 Hz.

Figure 6.

Second moment of area values for the ulna and the radius shown are measured from the proximal end. It can be seen that the variation is significant in some cases. The z-axis represents the proximal-distal direction and the x-axis the medial-lateral direction.

Figure 7.

Comparison of the average z-strain of selected elements on ulna (left) and radius (right) for a loading of 2 N @ 2 Hz. For each bone the the strain variations are shown from interosseous membrane stiffness of 2.5, 5, and 10 N/mm.

Figure 8.

Comparison between resultant force at forearm mid-shaft and applied load (top left: 3.5 N @ 0.2 Hz; top right: 3.5 N @ 2 Hz; bottom left: 2 N @ 0.2 Hz; bottom right: 2 N @ 2 Hz). In each figure the dotted line shows the applied load magnitude and the resultant force in the ulna and radius combined is shown for interosseous membrane stiffness of 2.5, 5, 10, and 15 N/mm.

Figure 9.

Compressive strain contour plots on the medial surface of the ulna and the radius from a FE simulation of 2 N peak load with a preload of 0.3 N and a interosseous membrane stiffness value of 10 N/mm. Contour plots are shown at times of 0.5, 1.0, 1.5, 2.0, and 2.5 s which represents one quarter cycle.

Figure 10.

Compressive strain contour plots on the medial surface of the ulna and the radius from a FE simulation of 2 N peak load with a preload of 0.3 N and a interosseous membrane stiffness value of 2.5, 5, 10 and 15 N/mm. Contour plots are shown at 2.5 s which represents the time at one quarter cycle.

Table 1

Strain Gauge Readings Under Various Preload Levels

Total load	Preload	Frequency	<u>Maximal Strain (Magnitude, $\mu\epsilon$)</u>	
			Ulna	Radius
2 N	0.3	2 Hz	-1830	-1790
	0.8	2 Hz	-1200	-1240
	0.3	0.2 Hz	-1810	-1910
	0.8	0.2 Hz	-1090	-1120
3.5 N	0.3	2 Hz	-3220	-3200
	0.8	2 Hz	-2673	-3030
	0.3	0.2 Hz	-3172	-3210
	0.8	0.2 Hz	-2420	-2430

Table 2Average Moment Of inertia (M.O.I), mm⁴

	Ulna (mm ⁴)		Radius (mm ⁴)	
	I _x	I _y	I _x	I _y
Average	0.01213	0.02232	0.01575	0.01986
Standard Deviation	0.00074	0.00806	0.00432	0.00512

Table 3
Comparison of Three-Point Bending Deflection Between Experimental and FE Results

	Load (N)	Displacement (mm)		Elastic Modulus (GPa)	
		Experimental	FE	Experimental Estimation	FE
Ulna (Initial load)	1.12	0.047	0.0468	13.8	14.3
Radius (Initial Load)	2.52	0.051	0.0509	9.9	9.4
				Maximum Compressive Strain ($\mu\epsilon$)	Maximum Tensile Strain ($\mu\epsilon$)
Ulna (Failure Load)	7.16	0.292	0.3257	23,480	19,600
Radius (Failure Load)	12.56	0.254	0.2627	48,000	31,170

Table 4

Summary of Peak Values of Average z-Strain Curves from all FE Simulations (Absolute Value, Units: Microstrain)

Axial Load		3.5 N			2 N		
Load Frequency		2 Hz		0.2 Hz		0.2 Hz	
Membrane Stiffness (N/mm)		Ulna	Radius	R/U	Ulna	Radius	R/U
2.5		8060	6248	0.78	6415	5719	0.89
5		5844	5263	0.90	5139	4828	0.94
10		4324	4198	0.97	3973	3970	1.00
15		3602	3521	0.98	3438	3481	1.01
Experimental Value		3220	3200	0.99	3172	3210	1.02
0.2 Hz/2 Hz Experimental Strain Ratio				0.985			0.997
							1.1

Note. All strain values are compressive strains. Experimental strain values are for the 0.3 N preload case.