Sex and Age Effects on Tibia Biomechanical Properties in Dynamic Four-Point Bending

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ABSTRACT

Globally, approximately 11 million pedestrian impacts each year result in injury or death. The lower extremity is the most frequently injured body region in pedestrian impacts, and these injuries often lead to long-term disability. To better understand lower extremity injury risk, biomechanical testing of the leg and tibia has previously been conducted. However, this work has mainly focused on males, despite females having a higher risk of sustaining lower extremity injuries in these loading events. This study aimed to evaluate the effects of sex and age on biomechanical properties of the isolated human tibia in a diverse sample. Sixty-six tibiae from 35 females and 31 males with similar age distributions ($p>0.05$; females=68.8±3.4 years; males=59.8±4.0 years) were utilized. Tibial diaphyses were loaded to failure at 6 m/s in lateral-medial four-point bending to replicate vehicle bumper interaction in pedestrian impacts. Reaction forces were measured at each end of the bone. Proximal and distal moments were calculated as the product of reaction force and the distance from the axis of rotation to the nearest impact site. Peak bending moment at midshaft was further calculated as the maximum of the average proximal and distal moments. Strain at time of fracture was determined from axial strain gages on the lateral and medial bone surface between impactor locations. Maximum deflection was normalized by specimen length and area moment of inertia. Males exhibited significantly greater peak moments than females ($p=0.001$). In the full sample, age demonstrated a weak but significant negative relationship with peak moment ($p=0.013$, $R^2=9.3\%$); however, no significant age relationship was observed when males or females were analyzed separately. Male tibiae had a significantly greater deflection than female tibiae ($p=0.001$), but no significant sex difference was observed with normalized deflection. Age was strongly and negatively associated with deflection in the full sample ($p<0.0001$, $R^2=39.7\%$), and for males ($p<0.0001$, $R^2=44.8\%$) and females ($p<0.0001$, $R^2=25.4\%$) separately. No significant sex differences were observed for tensile strain at fracture. However, age had a significant negative relationship with tensile strain in the full sample and within each sex ($p<0.0001$). Further investigation with larger sample sizes is needed to characterize the interacting effects of sex, age, and size. Overall, this work advances understanding of tibial biomechanics and provides insight for improving pedestrian safety tools including anthropomorphic test devices and human body models.

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INTRODUCTION

More than 11 million pedestrian-vehicle impacts result in injury or death worldwide each year (Network G.B.O.D.C., 2023). Pedestrian fatalities in the United States have increased by 78% since 2009, reversing decades of decline and returning to levels last seen in the late 1970s (Insurance Institute for Highway Safety, 2025). In 2023, an estimated 123,000 U.S. nonfatal pedestrian emergency department visits resulted from unintentional vehicle strikes, with average medical costs of $9,400 and quality-of-life losses of $73,000 per case (Centers for Disease Control and Prevention, 2023). The lower extremity is the second most commonly injured body region in pedestrian-vehicle impacts in the US (44%) (Mallory et al., 2024), and the most frequently injured region in France and Sweden (Saadé et al., 2020; Värnild et al., 2023). Multiple studies identify the tibia as the most commonly injured lower-extremity component in pedestrian impacts and the most frequent site of severe lower extremity trauma (Mizuno, 2005; Saadé et al., 2020). Tibia diaphyseal fractures are associated with substantial costs (Vanderkarr et al., 2023; Schade et al., 2021) as most require operative management (Wennergren et al., 2021). These injuries are also associated with notable complication rates, including infection, amputation, and nonunion, which further increase the overall cost of care (Schade et al., 2021).

The injury rates and severity of lower-extremity injuries vary across demographic groups. Older pedestrians have been shown to sustain more severe injuries than younger individuals (Saadé et al., 2020; Leo et al., 2021), and prior work has reported an increased risk of lower extremity injury with increasing age among impacted pedestrians (Värnild et al., 2023). Analysis of the Netherlands' BRON accident database (2000–2014) found female pedestrians were significantly more likely than males to sustain abbreviate injury scale (AIS) 2+ lower-extremity injuries in pedestrian-vehicle impacts (Leo et al., 2021). The analysis further highlighted the tibia as significantly more likely to sustain AIS2+ injury in pedestrian impacts of those ≥60 years and AIS3+ injuries in pedestrians impacts of those <60 years (Leo et al., 2021). However, these findings are not always consistent. The same study did not find these sex-differences to be significant across all databases and age groups (Leo et al., 2021), and other studies have found younger males to be significantly more likely to sustain tibial fractures in pedestrian impacts (Starnes et al., 2011). These discrepancies in observations highlight the need for a more comprehensive understanding of this complex loading scenario.

This study built on the work of Harden et al. (2023) to quantify human tibial biomechanical response using an experimental setup designed to simulate pedestrian impacts through realistic loading rates and four-point bending (Harden et al., 2021; Harden et al., 2023). The purpose of this study was to evaluate the effects of sex and age on biomechanical properties of the isolated human tibia in a diverse sample.

METHODS

Sixty-six tibiae from 35 females and 31 males with similar age distributions (Student's t-test, $p>0.05$; female=68.8 ± 3.4 years; male = 59.8 ± 4.0 years) were utilized in this study (Figure 1). Tibiae were acquired from donors of the Ohio State Body Donation Program (Columbus, Ohio,

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USA) in compliance with all ethics protocols. Only one randomly selected right or left tibia from each donor was included, and tibiae with any pre-existing trauma in the diaphysis were excluded from the sample.

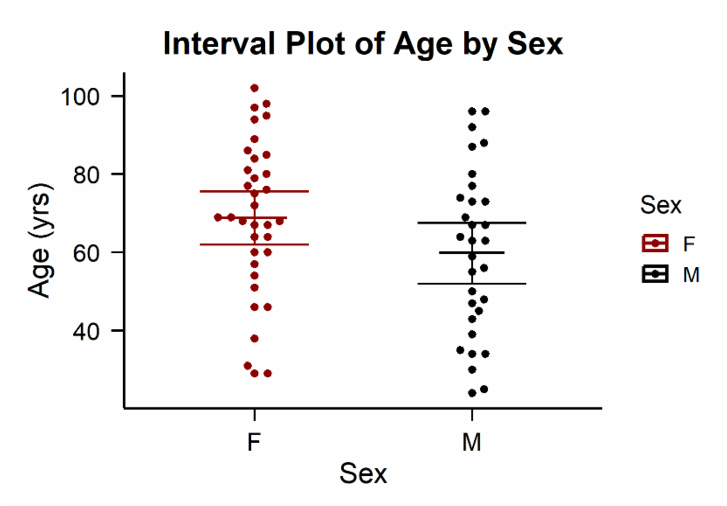

Figure 1. Sample Age Distributions by Sex (p>0.05)

Tibiae underwent a standardized preparation protocol prior to biomechanical testing. Excised tibiae were wrapped in saline-soaked gauze and stored at -20 °C following a storage protocol shown to have no significant effect on the mechanical properties of cortical bone (Linde et al., 1991; Reilly et al., 1974; Hamer et al., 1996). High-resolution computed tomography (HR-CT) scans were acquired for all tibiae with consistent acquisition parameters (120 kV, 262 mAs, 1024x1024 matrix, 0.335 mm in-plane resolution, Phillips Ingenuity 64-slice digital PET/CT).

Immediately prior to testing, all soft tissue except for the periosteum was removed. Each tibia was potted (Master Dyna-Cast Fast-Cast Urethane, Freeman Manufacturing and Supply Co., Avon, OH, USA) in a custom fixture at the 20% and 80% sites along the total bone length, excluding the medial malleolus, to ensure rigid, fully attached pots. An anatomically relevant coordinate system was utilized to ensure all tibiae were potted in the same orientation (Harden et al., 2021). Proximal and distal epiphyses were subsequently removed. Further details about the preparation process can be found in Figure 2 and Harden et al. (2021). Strain gages were applied to the lateral, medial, and posterior surfaces of the tibia diaphysis. Rosette gages were mounted at 55% of total bone length from the distal end (Figure 2). Only the axially aligned lateral and medial gages were included in this analysis.

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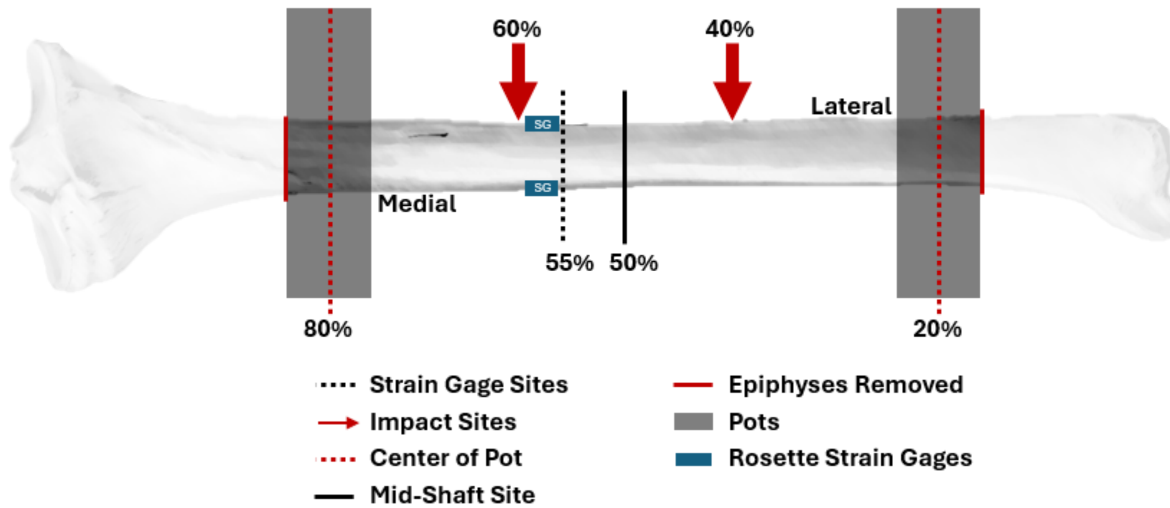

Figure 2. Tibia Schematic for Preparation and Data Collection

All tibiae were loaded in lateral-medial four-point bending at 6 m/s to simulate a vehicle-pedestrian blunt impact to the mid-diaphyseal region of the leg utilizing a custom-designed fixture mounted to a custom-built material testing system (MTS) (High-Rate Material Testing System, MTS Systems Corporation, Eden Prairie, MN, USA). The set-up was designed to simultaneously impact the 40% and 60% locations along the total bone length at a constant rate by using an adjustable impactor fixture mounted on the actuator (Ebacher et al., 2007; Harden et al., 2021) (Figure 3A). The test fixture was rigidly mounted at the center of the base of the MTS and consisted of two potting assemblies mounted to a slotted baseplate (Figure 3B), allowing the span to be adjusted. Each assembly included a connector plate secured to the baseplate (Figure 3C), a load cell mounted above it (Figure 3D), and an upper subassembly that tightly secured the potted bone end (Figure 3E). The coordinate system used for testing and data processing corresponds to the SAE J211 Instrumentation for Impact Test Standard (SAE International, 2022). Rotary bearings permitted rotation about the X-axis and translation along the Z-direction (Figure 3). This design reduced moments at the bone ends during bending, permitting free bending.

Six-axis load cells (Bertec Corporation, Columbus, OH, USA) were positioned beneath the potting assemblies. The MTS provided actuator displacement at 20 kHz, and all external instrumentation was routed to an external high-rate data acquisition system (DTS SLICE PRO, Seal Beach, CA, USA) for data collection at 100 kHz. Load data were filtered at a 4 kHz cutoff frequency using a post-processing phase-less 4th-order low-pass Butterworth filter. MTS displacement and strain were not filtered. From pre-test HR-CT scans, area moment of inertia in the lateral-medial orientation was quantified using Dragonfly software (Dragonfly 3D World V2025.1, Comet Technologies Canada Inc, Montréal, Canada) for one slice at the midshaft (50%) location. The start, duration, and fracture time of the impact event was determined from strain data obtained from the axially aligned gage of the rosettes on the bending surface.

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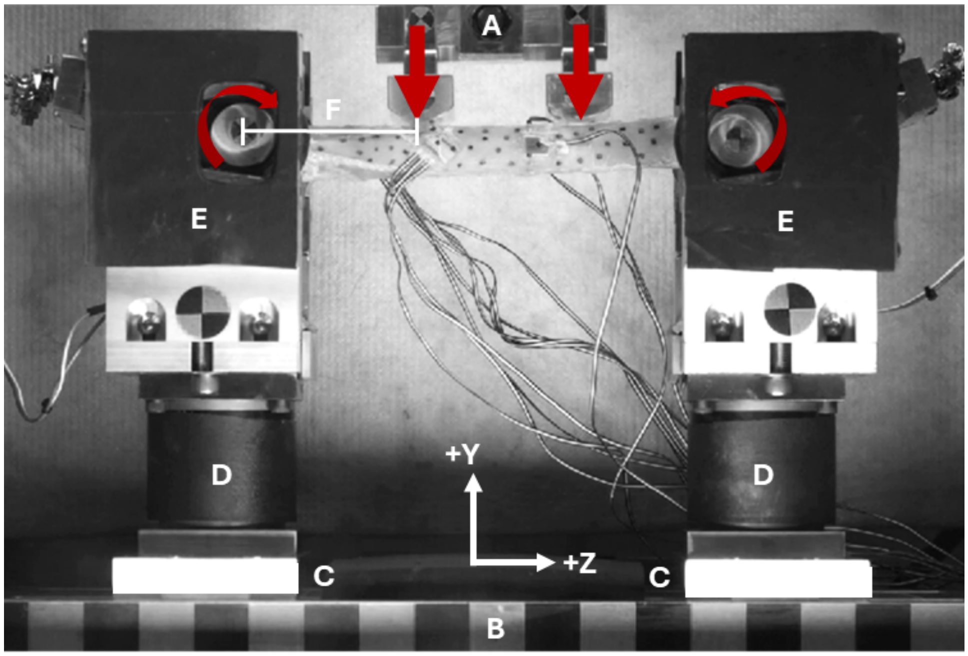

Figure 3. Tibia Test Set-Up Overview (A: Adjustable impactor fixture mounted on the actuator, B: Slotted baseplate, C: Connector plates, D: Load cells, E: Upper subassembly that secures the potted end of the bone and allows for Z-axis translation, F: Moment arm)

All tibia biomechanical properties are defined in Table 1. Peak force was calculated as the maximum summed force recorded across the time history from both load cells in the primary loading direction (Figure 3). Proximal and distal bending moments were calculated by multiplying the force measured from each load cell by a moment arm, defined as the distance between its corresponding pot axis of rotation and nearest impactor (Figure 3F). Mid-shaft bending moments were calculated as the average of the proximal and distal bending moments, representing the bone center in shear–moment analysis. The maximum value of this mid-shaft moment over the time history was defined as the peak bending moment.

Tibial deflection was defined as the product of the event duration and the MTS displacement rate in the primary loading direction. Deflection values (δ) were normalized by a ratio of the cubed span length (L), defined as the distance between the 20% and 80% sites, over the area moment of inertia (I) at the 50% site to account for differences in tibia size (Eq. 1).

$$\delta_{Normalized} = \delta \frac{L^3}{I} \qquad \text{[Eq. 1]}$$

The strain magnitude at fracture initiation defined compressive (lateral surface) and tensile (medial surface) strains. Due to the higher frequency of tensile gage failure, event durations were determined using the gage on the compressive surface.

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JMP statistical software (Student Edition 18) and a significance level of $\alpha = 0.05$ was used for all statistical analyses. Two-tailed Student's t-tests were conducted to compare biomechanical properties between sexes. Pearson correlations were performed to investigate the relationships between cross-sectional geometry variables at the midshaft. Correlations were categorized as weak (<0.5), moderate (0.5–0.8), and strong (>0.8). Univariate linear regression analyses were performed to assess relationships between biomechanical properties and age. Analysis of Covariance (ANCOVA) models were used to analyze the potential interaction effects of sex and age on tibial biomechanical response.

All statistical models were evaluated for normality, model adequacy, and the presence of outliers. Normality was assessed using normal quantile (Q–Q) plots of the residuals and the Shapiro–Wilk goodness-of-fit test on the residual distribution. Although the Shapiro–Wilk test did not always support the assumption of normality, the sample size met the requirements of the Central Limit Theorem. In most cases, departures from normality were attributable to known outliers, which were retained in the statistical analyses. Model adequacy was evaluated by examining residuals by row and residuals versus predicted values. Outliers were identified using outlier box plots of the residual and Studentized residual distributions. Prior to all Student's t-tests, equality of variances was assessed; a pooled t-test was used if variances were equal, and a separate-variance t-test if they were unequal. Variables used in statistical analysis and their definitions can be viewed in Table 1.

Table 1. Variables and Definitions

Demographics	
Sex	Biological classification as Male or Female
Age (years)	Age at death
Bone Properties	
Mechanical Span (L, mm)	Distance between 20% and 80% sites (i.e., diaphysis)
Area Moment of Inertia (I, mm^4)	Measure of resistance to bending and deflection
Structural Properties	
Peak Force (N)	Sum of maximum force from each load cell
Peak Bending Moment (Nm)	Maximum moment at mid-diaphysis
Deflection (mm)	Displacement to fracture
Normalized Deflection	Deflection multiplied by L^3/I
Tensile Strain (με)	Strain at fracture from medial strain gage
Compressive Strain (με)	Strain at fracture from lateral strain gage

RESULTS

Descriptive and test statistics comparing males to females for age, mechanical span, and cross-sectional geometry, are provided in Table 2. Males had a significantly larger mechanical span and greater area moment of inertia than females ($p \leq 0.0001$). Descriptive and test statistics comparing

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males to females for all biomechanical variables are provided in Table 3. Several tests did not yield an event duration due to strain gage failure during impact, and tibial deflection could not be computed for these specimens, resulting in reduced sample sizes for some variables. Both the full sample and the sex-specific subgroups in the reduced samples were normally distributed with no significant age differences between sexes ($p>0.05$). Mechanical span exhibited no significant relationship with peak force or peak bending moment in the full sample ($p=0.636$, $p=0.090$, respectively). However, mechanical span in the male sample had a significant relationship with peak force ($p=0.0482$, $R^2=13.2\%$). Mechanical span exhibited no significant relationship with peak force in females or peak bending moment in either sex ($p>0.05$).

Table 2. Age, Span, and Area Moment of Inertia by Sex

	Sex	*N*	*Min*	*Max*	*Mean*	*SD*	*p-value*
Age (yrs)	M	31	24	96	60	21	0.0786
	F	35	29	102	69	20	
Mechanical Span (mm)	M	31	201	248	229.03	12.11	**< 0.0001**
	F	35	173	237	208.69	13.62	
Area Moment of Inertia (mm^4)	M	31	6980	21322	12875	518	**< 0.0001**
	F	35	4598	15234	7610	488	

Significant p-values are **bold**

No significant sex differences were observed for peak force ($p=0.056$) (Table 3). Age had a significant negative relationship with peak force in the full sample ($p=0.037$) but not for males ($p=0.313$) or females ($p=0.159$) using simple regressions (Figure 4, Table 4). An ANCOVA model found significance in the whole model ($p=0.036$, $R^2=7.2\%$), but no significant contributions by sex ($p=0.127$) or age ($p=0.083$) individually. However, when including an interaction term between sex and age, the model was no longer significant for peak force ($p=0.079$).

Males exhibited significantly greater peak moments than females ($p=0.001$) (Table 3). In the full sample, age demonstrated a weak but significant negative relationship with peak moment ($p=0.013$, $R^2=7.9\%$); however, no significant age relationship was observed when males ($p=0.290$) and females ($p=0.097$) were analyzed separately (Figure 5, Table 4). An ANCOVA model ($p=0.001$, $R^2=18.7\%$) confirmed a significant effect of sex ($p=0.003$) on peak moment after controlling for age.

Table 3. Biomechanical Properties by Sex

	Sex	*N*	*Min*	*Max*	*Mean*	*SD*	*p-value*
Peak Force (kN)	M	31	7.9	25.4	16.8	3.7	*0.0559*
	F	35	4.2	22.3	14.8	4.4	
Peak Bending Moment (Nm)	M	31	334.5	1023.5	671.2	139.9	***0.0008***
	F	35	141.9	804.6	542.4	155.6	
Deflection (mm)	M	30	3.2	10.6	6.8	1.8	***0.0013***
	F	30	2.4	7.9	5.3	1.5	
Normalized Deflection	M	30	2730	13946	6765	500.35	*0.8437*
	F	30	1248	12964	6625	500.35	
Tensile Strain (με)	M	25	3639	14813	8569	3091	*0.1330*
	F	33	2909	11767	7446	2518	
Compressive Strain (με)	M	30	2459	11329	7433	2192	*0.1173*
	F	30	1312	9922	6489	2403	

Significant p-values are **bold**

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Table 4. Relationships of Sex and Age with Force and Moment

Statistical Test	Independent Variable(s)	Peak Force			Peak Bending Moment		
		p	*Adj. R^2*	*Robust p*	*p*	*Adj. R^2*	*Robust p*
Linear Fit	**Age**	**0.0370**	5.2%	**0.020**	**0.0127**	7.9%	**0.0042**
	Male	0.313	0.2%	0.385	0.290	0.5%	0.2890
	Female	0.159	3.1%	0.074	0.097	5.3%	0.060
ANCOVA	**Age, Sex, Sex*Age**	0.079	6.0%	-	**0.0017**	17.7%	-
	Age	0.084	-	-	0.052	-	-
	Sex	0.128	-	-	**0.0032**	-	-
	*Age*Sex*	0.669	-	-	0.600	-	-
	Age & Sex	**0.0357**	7.2%	-	**0.0006**	18.7%	-
	Age	0.083	-	-	0.051	-	-
	Sex	0.127	-	-	**0.0031**	-	-

Significant p-values are **bold**

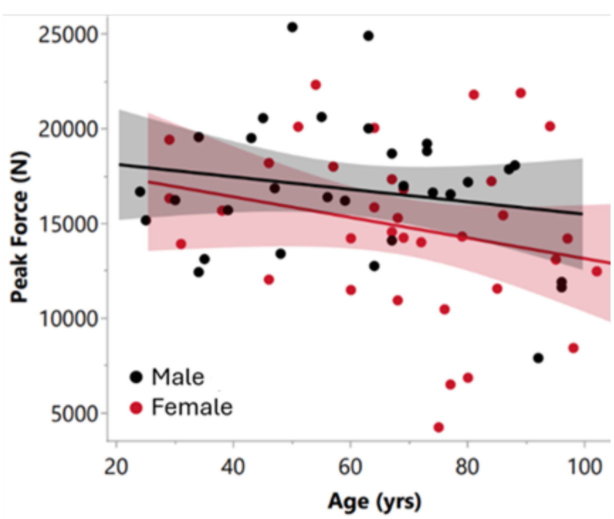

Figure 4. Peak Force with Age

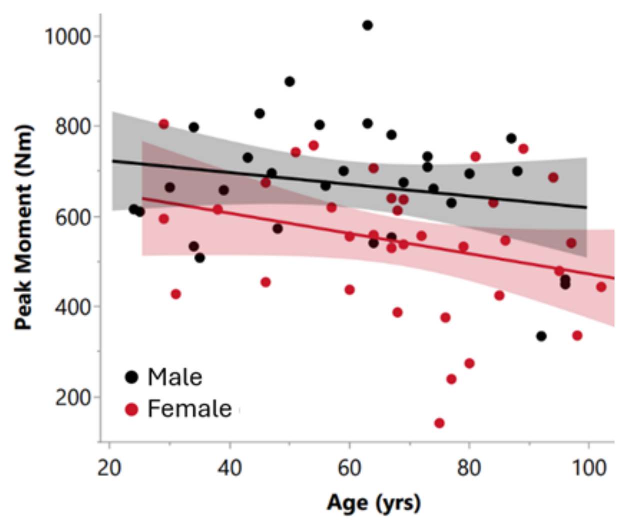

Figure 5. Peak Bending Moment with Age

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Male tibiae had a significantly greater deflection than female tibiae (p=0.001) (Table 3). Age was strongly and negatively associated with deflection in the full sample (p<0.0001, R^2=39.7%), and for males (p<0.0001, R^2=44.8%) and females (p=0.003, R^2=25.4%) separately (Figure 6, Table 5). An ANCOVA model (p<0.0001, R^2=46.1%) indicated that deflection was significantly influenced by age (p<0.0001) and sex (p=0.007), but the model exhibited a significant lack of fit (p<0.05).

No significant sex differences were observed for normalized deflection (p=0.844) (Table 3). Age was negatively associated with normalized deflection in the full sample (p=0.009, R^2=9.7%), and for males (p=0.006, R^2=21.6%) but not for females (p=0.412) separately (Figure 7, Table 5). An ANCOVA model (p=0.0312, R^2=8.4%) indicated that normalized deflection was significantly influenced by age (p=0.009) but not sex (p=0.677), and the model exhibited a significant lack of fit (p<0.05).

Table 5. Relationships of Sex and Age with Deflection

Statistical Test	Independent Variable(s)	Deflection (mm)			Normalized Deflection		
		p	*Adj. R^2*	*Robust p*	*p*	*Adj. R^2*	*Robust p*
Linear Fit	**Age**	**<0.0001**	39.7%	**<0.0001**	**0.0090**	9.7%	**0.0069**
	Male	**<0.0001**	44.8%	**<0.0001**	**0.0056**	21.6%	**0.0006**
	Female	**<0.0001**	25.4%	**<0.0001**	0.412	0%	0.367
ANCOVA	**Age, Sex, Sex*Age**	**<0.0001***	46.5%	-	**0.0286**	10.2%	-
	Age	**<0.0001**	-	-	**0.0096**	-	-
	Sex	**0.0068**	-	-	0.683	-	-
	*Age*Sex*	0.232	-	-	0.144	-	-
	Age & Sex	**<0.0001***	46.1%	-	**0.0312***	8.4%	-
	Age	**<0.0001**	-	-	**0.0089**	-	-
	Sex	**0.0071**	-	-	0.677	-	-

Significant p-values are **bold**
*ANOVA tests with significant lack of fit

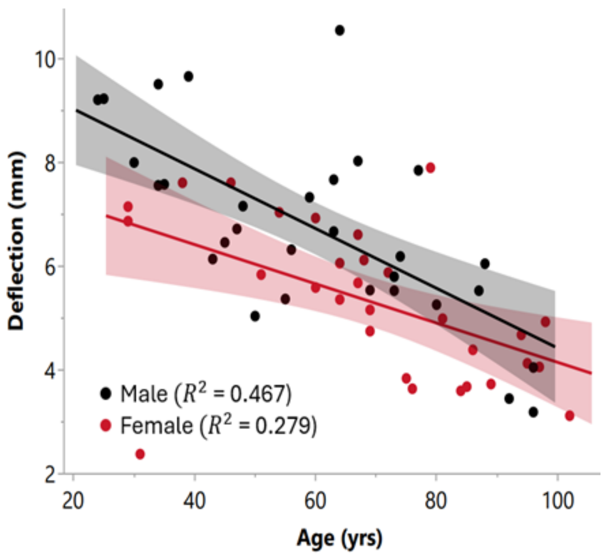

Figure 6. Deflection with Age

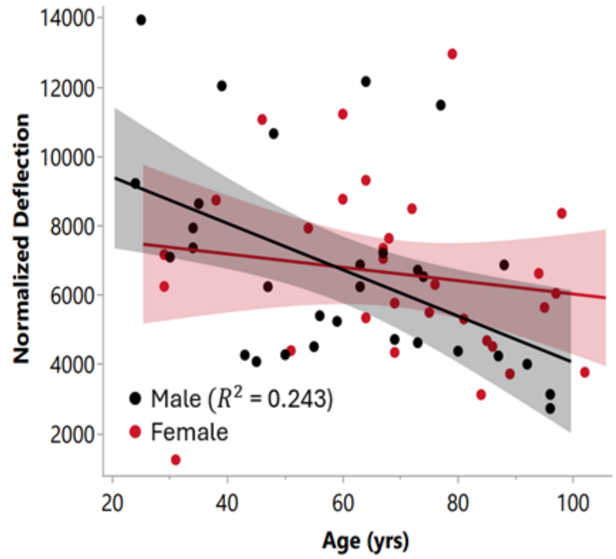

Figure 7. Normalized Deflection with Age

No significant sex differences were observed for tensile strain at fracture (p=0.133) (Table 3). However, age had a significant negative relationship with tensile strain in the full sample (p<0.0001, R^2=36.9%), and for males (p<0.0001, R^2=47.3%) and females (p=0.003, R^2=23.3%) separately (Figure 8, Table 6). An ANCOVA model (p<0.0001, R^2=36.6%) confirmed a significant effect of age (p<0.0001) but not sex (p=0.385) on tensile strain. No significant sex differences were observed for compressive strain at fracture (p=0.117) (Table 3). However, age had a significant negative relationship with compressive strain in the full sample (p<0.0001, R^2=37.1%), and for males (p<0.0001, R^2=42.6%) and females (p=0.002, R^2=26.4%) separately (Figure 9, Table 6). An ANCOVA model (p<0.0001, R^2=36.5%) confirmed a significant effect of age (p<0.0001) but not sex (p=0.538) on compressive strain.

Table 6. Relationships of Sex and Age with Strain

Statistical Test	Independent Variable(s)	Tensile Strain			Compressive Strain		
		p	*Adj. R^2*	*Robust p*	*p*	*Adj. R^2*	*Robust p*
Linear Fit	**Age**	**<0.0001**	36.9%	**<0.0001**	**<0.0001**	37.1%	**<0.0001**
	Male	**<0.0001**	47.3%	**<0.0001**	**<0.0001**	42.6%	**<0.0001**
	Female	**0.0026**	23.3%	**0.001**	**0.0022**	26.4%	**0.0012**
ANCOVA	**Age, Sex, Sex*Age**	**<0.0001**	37.7%	-	**<0.0001**	35.4%	-
	Age	**<0.0001**	-	-	**<0.0001**	-	-
	Sex	0.413	-	-	0.540	-	-
	*Age*Sex*	0.171	-	-	0.803	-	-
	Age & Sex	**<0.0001**	36.6%	-	**<0.0001**	36.5%	-
	Age	**<0.0001**	-	-	**<0.0001**	-	-
	Sex	0.385	-	-	0.538	-	-

Significant p-values are **bold**
*ANOVA tests with significant lack of fit

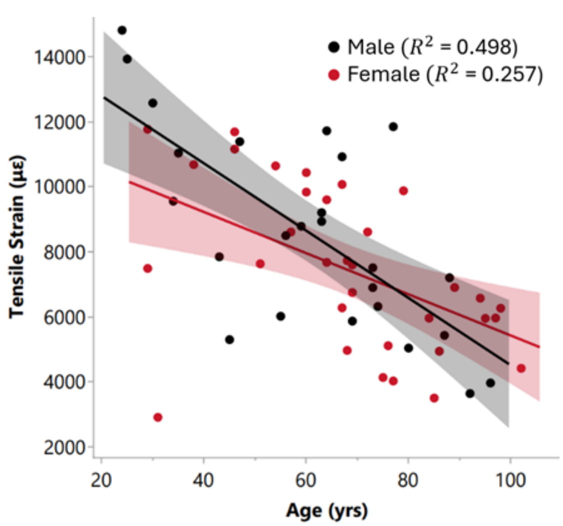

Figure 8. Tensile Strain with Age

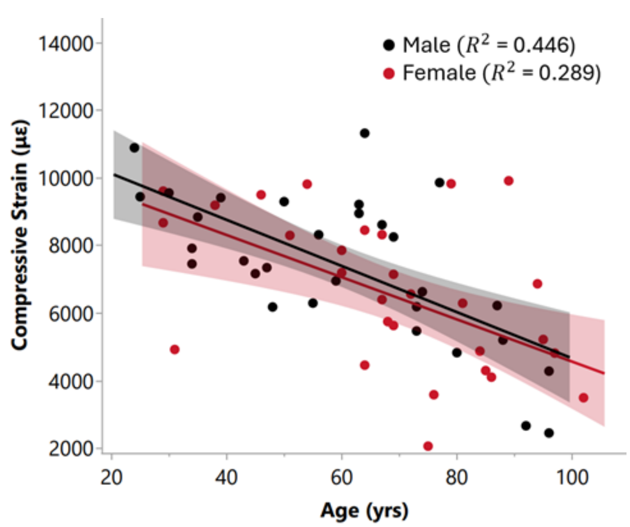

Figure 9. Compressive Strain with Age

DISCUSSION

Loading Observations

Although a weak overall age-related trend was observed in the variables used to quantify tibial loading, sex-specific age-related trends were not evident. The finding that sex differences emerged for peak moment, but not peak force, suggests that accounting for size-related factors is important for detecting sex-related differences in tibial loading. Peak force and bending moment decreased with age in the full sample, but no significant trends were observed in sex-specific analyses, indicating that the underlying relationship with age is more nuanced (Table 3, Table 4). The absence of significant relationships in these biomechanical properties when isolated by sex is likely due to reduced sample sizes within each group, as no outliers were identified in either analysis. When evaluating sex without age, significant differences in peak moments but not peak force were observed, suggesting observed sex differences in peak moment are primarily attributable to the added factor of length (Table 3). Although peak moment accounts for variations in bone length in characterizing loading, the observed differences suggest that additional factors, such as cross-section geometry, shape, and microstructure, may contribute.

Several previous studies have conducted comparable tests on the tibia in dynamic bending, though all have focused specifically on quantifying the mechanical response of the human tibia in dynamic three-point bending (Mather et al., 1968; Nyquist et al., 1985; Schreiber et al., 1998; Kerrigan et al., 2003; Kerrigan et al., 2004; Bing, 2011; Cameron et al., 2020; Hardy et al., 2024). However, most of these studies have primarily focused on quantifying the overall biomechanical behavior rather than evaluating demographic differences. Table 7 summarizes the test conditions and results from studies most comparable to the current work. It should be noted that the summarized results combine tests with differing loading directions and tissue conditions within the same study.

Table 7. Dynamic Tibia Bending Studies from Literature vs. Current Study

Study	Sample	Loading	Condition	Component*	Average Loading Rate (m/s)	Average Failure Force (kN)	Average Failure Bending Moment (Nm)
Nyquist 1985	15 M 4 F	3Pt	Fleshed	Full	3.83 M 2.25 F	4.72 M 4.29 F	308.0 M 272.3 F
Schreiber 1998	6 M 6 F	3Pt	Fleshed	Full	5.55	NA	417.0 M 369.8 F
Kerrigan 2003	4 M	3Pt	3 Fleshed 1 Denuded	Full	1.45	3.45	295.7
Kerrigan 2004	4 M	3Pt	Fleshed	Full	1.56	3.71	363.0
Bing 2011	6 M	3Pt	3 Fleshed 3 Denuded	Full	1.29	3.42	303.2
Cameron 2020	4 M 2 F	3Pt	Denuded	Full	8.33	5.41 M 3.85 F	345.3 M 243.5 F
Hardy 2024	7 M 9 F	3Pt	Denuded	Diaphysis	5.00	6.02 M 4.15 F	NA
Current Study	31 M 35 F	4Pt	Denuded	Diaphysis	6.00	16.82 M 14.85 F	671.2 M 542.4 F

*Includes which studies potted epiphyses (Full) as opposed to only the diaphysis (Diaphysis)

Previous studies investigating tibial response under dynamic bending have generally been limited by small sample sizes, inconsistent loading conditions, and limited representation across sex. Most prior work did not directly evaluate the influence of age on biomechanical response, leaving important gaps in understanding how sex and age contribute to tibial injury mechanisms. Previous dynamic bending studies of the tibia have utilized three-point bending; however, four-point bending more accurately represents tibial loading during a vehicle-pedestrian side impact and was therefore better suited for this study.

Comparison of the results to previous work must be done cautiously due to the differences in the test-set ups. While differences in the bending direction between studies have negligible effects (Nyquist et al., 1986), the presence of flesh, bending configurations, loading rates, and potted regions were inconsistent across studies and must be considered when making comparisons. Previous studies have hypothesized that the presence of the surrounding flesh and fibula increases

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the energy to fracture a leg in bending (Kerrigan et al., 2003; Bing, 2011); However, these suggested effects were based on limited sample sizes, and therefore, the contribution of these components to the tibial biomechanical response remains unclear. It is also important to consider the differences in loading rate, as the mechanical response of cortical bone has been proven to depend on the loading rate (McElhaney et al., 1966; Kemper et al. 2007). Furthermore, the points of rotation and support differed between set-ups, as variations in the potting region affects the mechanical response. Most previous studies potted the tibial epiphyses (Schreiber et al., 1998; Kerrigan et al., 2003; Kerrigan et al., 2004; Bing et al., 2011; Cameron et al., 2020), which increases the length of the tested specimen. The current study potted the tibia at consistent relative locations based on total length to effectively isolate diaphyseal cortical shaft behavior. Due to these variations in test set ups, it is difficult to draw meaningful comparisons between the current study and most previous work.

The most comparable study to compare three-point bending to four-point bending is by Hardy et al. (2024) since the study tested at a loading rate (5 m/s) similar to the current study (6 m/s), excluded the fibula and surrounding soft tissue, and removed the epiphyses (Hardy et al., 2024). The average peak forces observed in Hardy et al. (2024) were greater than the previous studies listed in Table 7 but less than the current study. The observed greater peak force in Hardy et al. (2024) compared to prior three-point bending studies suggests that epiphyseal potting may influence the mechanical response under loading, and despite its anatomical relevance in real-world fractures, the inclusion of the epiphyses introduces an additional source of variability in the experimental set-up. This increased force observed in the current study compared to the study by Hardy et al. (2024) suggests an additional inertial effect in four-point bending. When bending at higher loading rates, regardless of bending set-up, the impactor must accelerate the mass beneath it over a shorter time interval, thereby increasing inertial effects. In four-point bending, this effect is further amplified due to the larger mass engaged beneath the impactor, resulting in increased inertial contributions.

Despite differences from previous bending studies, the current study effectively represents vehicle bumper–pedestrian loading of the diaphysis and provides useful insight into the loading experienced by the tibia during these interactions. Furthermore, the results indicate that higher loads than previously reported may occur in pedestrian-vehicle bumper impacts when accounting for the effects of a distributed load.

Deflection and Strain Observations

Age strongly influenced deformation-related response metrics. No significant sex differences were observed in normalized deflection, compressive strain, or tensile strain (Table 3). This indicates similar bone deformation measures between sexes, and sex differences observed in other biomechanical properties are possibly due to geometrical and cross-section differences. Deflection and strain significantly decreased with age across both full and sex-specific samples (Table 5, Table 6), suggesting increased tibial brittleness and stiffness with age. Previous studies have reported similar findings, demonstrating increased brittleness with age in cortical bone (Zioupos et al., 2020). Therefore, older populations may be at increased risk of sustaining AIS 2+

injuries in pedestrian-vehicle crashes due to a reduced capacity to dissipate energy through flexion and a greater propensity for brittle fracture.

However, the deflection results changed once normalized by length and area moment of inertia, with age only predicting normalized deflection in males (Table 5). Furthermore, the strength of the relationships decreases substantially upon normalization. When removing the outliers identified by the model residual distribution from the female normalized deflection data, age displayed a significant negative trend with normalized deflection as well. Thus, a contribution to the observed differences in results between deflection and normalized deflection is due to the presence of outliers retained in the female normalized deflection data in combination with reduced sample size when separated by sex. However, the dramatic decrease in the strength of this relationship suggests additional factors, such as the cross-section, gross shape, or microstructure, may play a role. Nevertheless, the observed outliers represent valid data points and were retained in the analysis to fully capture human variability.

Additional data is needed to reduce the influence of outliers, rather than removing them entirely, to draw more robust conclusions about how normalized deflection changes with age. However, the results generally support the notion that strain prior to fracture decreases with age, indicating increased tibial brittleness with age and consistent with previous literature (Zioupos et al., 2020).

Limitations

There were several limitations in this study that have not yet been discussed. The presence of outliers were retained in all statistical analysis to fully capture the observed human variation. Bending duration was determined from compressive-surface strain gages due to frequent tensile gage failure, although no significant duration differences were observed between gages. Male and female specimens were not size matched, with males exhibiting longer tibiae on average. However, these data still represent real human variation of bone sizes between males and females and the complicated relationship between sex and size. Additionally, the four-point bending setup did not use swiveling anvils, introducing the possibility of slight asymmetry in initial impactor contact.

Despite these limitations, the data presented still well-represents the interaction of a pedestrian-vehicle bumper interaction more accurately than prior three-point bending studies, and the data can be used to improve anthropomorphic test devices (ATDs) and human body models (HBMs) aimed to represent the pedestrian in side impacts when the vehicle bumper impacts the leg near the center of the tibia bone.

CONCLUSIONS

Age more strongly influenced deflection-related biomechanical response metrics, while sex differences were most apparent in peak bending moment and unnormalized deflection, both of which are influenced by tibial length. There was no significant interaction between sex and age for any biomechanical response; however, differing age-related trends were observed between sexes, indicating the need for additional data to better assess a potential interaction.

The significantly lower observed peak moments in females may explain why females sustain higher injury rates in prior epidemiological studies of pedestrian-vehicle impacts. Furthermore, the significant decreases in deflection, normalized deflection, and strain with increasing age support epidemiological evidence indicating that injury risk increases with age. Understanding sex-specific and age-related differences is useful in establishing injury risk, as well as improving female ATDs and (HBMs). Additionally, the dataset supports the development and validation of finite element models, enabling more accurate estimation of mechanical response across sex, size, and age without the confounding factors of soft tissue and trabecular bone. These models can further inform material property assignments and enhance the biofidelity of HBMs. Future work will aim to assess cross-sectional influence on biomechanical properties and incorporate testing of tibiae with intact soft tissue and fibular support, enabling direct comparison with denuded tibia models and prior experimental studies. Overall, these findings highlight the critical differences in injury mechanisms across sex and age and their implications for understanding fracture risk in pedestrian impacts.

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