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Strain-Based Methodology To Assess The Load Transmitted Into The Clavicle During Lateral Shoulder Impacts

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ABSTRACT

During car side impacts, the shoulder, especially the clavicle, is often injured. Determining the load transferred through the clavicle in situ during a shoulder lateral impact would allow a better understanding of shoulder injury mechanisms and of the role of the clavicle as a load path in this impact mode. The goal of this study was to develop a minimally invasive methodology to estimate the load transmitted into the clavicle during shoulder lateral impacts by measuring in situ strains.

Ten clavicles were harvested from six post-mortem human subjects (PMHS) and tested ex situ in compression with a device designed to mimic the anatomical boundary conditions in the shoulder. The compression was performed by imposing lateral and oblique displacement (up to ±25° off-axis) to the lateral extremity of the bone. The resulting forces and moments were measured as well as the strains by means of unidirectional strain gages pasted at different locations on the bone.

A linear relationship was observed between force and strain, regardless of the direction of the imposed displacement, and the angle of the imposed displacement was found to have almost no effect on the direction of the resulting force vector or on the slope of force-strain linear relationship. Based on these results, an empirical methodology is proposed to estimate the load transferred to the clavicle using in situ strain measurements.

INTRODUCTION

Shoulder injuries are a frequent result of automotive side impacts, and the clavicle provides an important load path in these exposures. This is apparent in the injury outcomes, as pointed out by Frampton et al. (1997) who reported that 66% of shoulder injuries during car lateral impacts are clavicle fractures. Hynd et al. (2007) highlighted the significant magnitude of upper limb trauma (arm and shoulder) in a retrospective study based on medical notes, and listed clavicle fracture as the most frequent injury in side impacts. Melvin et al. (1998) suggested that loading the shoulder rather than the thorax could avoid severe injuries to the internal organs by minimizing chest deflection. Thus, a better understanding of the force paths into and

through the shoulder could guide the design of safety devices to improve the protection of both the shoulder and the thorax during car side impacts.

Several experimental studies (Compigne et al., 2004; Bolte et al., 2000; Koh et al., 2005) have focused on the shoulder response to lateral impacts and, consistent with the field data, the main shoulder injury observed during these tests was clavicle fracture. None of the existing studies, however, have attempted to quantify the amount of the total applied force that is transmitted through the clavicle. Compression tests on isolated clavicles have been performed (Harnroonroij et al., 2000; Duprey et al., 2007), so the sub-failure structural behavior, as well as the force and displacement failure thresholds are reasonably well understood ex situ. There has been no attempt, however, to isolate the role of the clavicle as part of the shoulder structure in situ.

Determining the load transferred to the clavicle during a shoulder lateral impact would allow a better understanding of the shoulder loading process. The use of a load cell for nominally in situ measurements on the clavicle of post-mortem human subjects (PMHS) would significantly alter and damage the shoulder structure, and undoubtedly increase the risk of artifactual injuries, whereas several uniaxial strain gages can be placed on the in situ clavicle with minimal disruption of the load-bearing structures. For some long bones, the internal loads and moments generated by external forces can be calculated based on strain measurements by applying linear beam theory if the loads remain in the linear elastic regime. Previous investigators have used this method to quantify the sub-failure loading of long bones of live animals (Rybicki et al., 1977) and human cadavers (Funk et al., 2006). There are many simplifications associated with the application of beam theory in this context, and the range of applicability of the method is limited. For example, this method has been found to be reasonably accurate for pure flexion, but not for axial loading. Funk et al. (2006) observed that the error between the force calculated using the beam theory and the true force could be as much as 50%, for the tibia under pure compression. They attributed this large difference to bone curvature and deviations from assumptions of bone homogeneity and linearity. Therefore, the utility of this method for compressive loading of the clavicle is suspect even without consideration of the clavicle's complex geometry relative to the long bones considered in the past. The clavicle's shaft, which exhibits a double curvature producing an S-shape, further complicates the application of the beam theory and likely excludes this as a viable approach. Thus, applying the beam theory to the clavicle bone was not considered in the present study. The focus was put on establishing a primarily empirical methodology to determine the load transferred to the clavicle from strain measurements.

METHODS

Bone preparation

Ten clavicles were harvested from six subjects (4 males, 2 females, average age: 62 years, average size and weight: 1.75 m, 86 kg) in accordance with ethical guidelines and research protocols approved by the University of Virginia oversight committee. After all soft tissues were removed from the bones, the extremities of each bone were potted with an epoxy resin (FastCast®) in cylindrical cups (3.8 cm-diameter, 2.5 cm-depth). A potting jig was used to ensure that the 2 cups were parallel and collinear. Unidirectional strain gages (Vishay measurements, CEA-06-062-UW-350) were pasted on the bones at various locations: on the superior surface and around the cross-section at the level of the maximum of the posterior concavity.

Set-up

The specimens were horizontally mounted in a linear displacement stage which imposed lateral or oblique displacement at quasi static rates (0.63 and 20 mm/s) to the acromial extremity of the bone (Figure 1). At the sternal extremity a pin joint allows the clavicle to rotate around the y-axis; furthermore, the jig was also designed to allow or lock the z-rotation in order to evaluate the contribution of this degree of freedom (Figure 1).
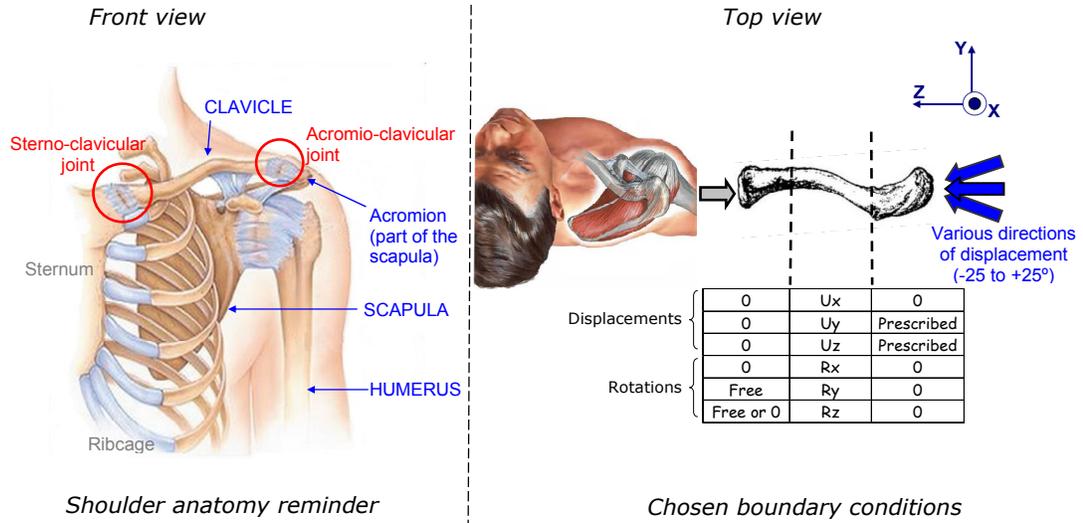
Figure 1:   Shoulder anatomy and boundary conditions chosen for testing.

A 6-axis load cell was placed at the acromial extremity of the clavicle in order to measure the forces and moments generated by the applied displacements. A potentiometer measured the magnitude of the input displacement at the distal end of the bone, and the gages provided strain measurements.

Compression tests

Each specimen was subjected to a battery of tests with (1) displacement applied at an angle varying from 0º to ±25º, (2) 2 quasi-static velocities, and (3) free or locked z-rotation (Table 1). Selected conditions were repeated to assess the repeatability of the methodology.

Table 1.   Matrix test, number of clavicles tested in each configuration.

	Low velocity * Axial rotation=Free	Low velocity * Axial rotation=locked	Higher velocity ** Axial rotation=Free	Higher velocity ** Axial rotation=locked
-25º	2	1		
-20º	2	1		
-15º	8	7	5	3
-10º	2	1		
-5º	2	1		
0º	10	9	6	6
+5º	2	1		
+10º	2	1		
+15º	8	7	5	3
+20º	2	1		
+25º	2	1		

*=0.63mm/s, **=20mm/s

RESULTS

Good repeatability of the tests was observed (Figure 2) when performing tests with similar conditions on the same bone at different times (before and after a battery of tests). This shows that the fixture was able to produce repeatable loadings, and that the battery of quasi-static tests performed in this study does not damage the bone.
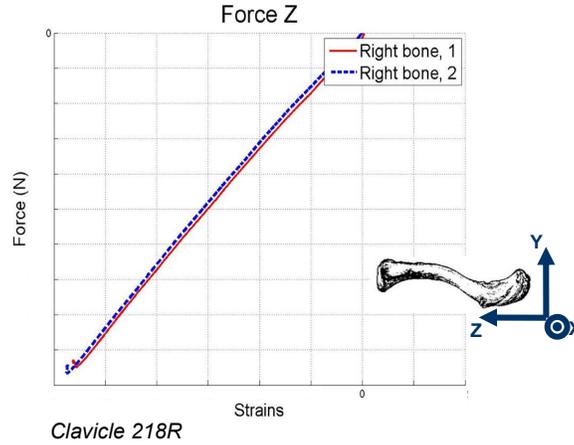
Clavicle 218R

Figure 2: Repeatability of the tests: force F_Z versus strains (same test conditions for 2 different tests, for the clavicle #218R).

The resulting moments were small. Table 2 indicates the maximum values obtained for all the clavicles, for the low-velocity tests, with free z-rotation, at 0º and at the maximum angles (+25º and -25º).

Table 2: Maximum moments (absolute values) reached over all the clavicles (tests at lower velocity, free z-rotation).

	0º	±25º
Mx	≤ 4 Nm	≤ 3 Nm
My	≤ 5 Nm	≤ 0.5 Nm
Mz	≤ 0.25 Nm	≤ 0.3 Nm

No significant difference was observed for force and strain results between tests performed with locked or free z-rotation. As anticipated, the moment around the Z axis depends on this boundary condition for all the tested directions of loading (0º, +15º and -15º, Figure 3). However, the maximum moments reached were very small (≤0.3Nm) and it seemed reasonable to consider that locking this rotation does not engender significant changes. This result indicates that the compression of a clavicle does not create significant torsion in the bone, at least not over the limited displacement input magnitudes considered here.
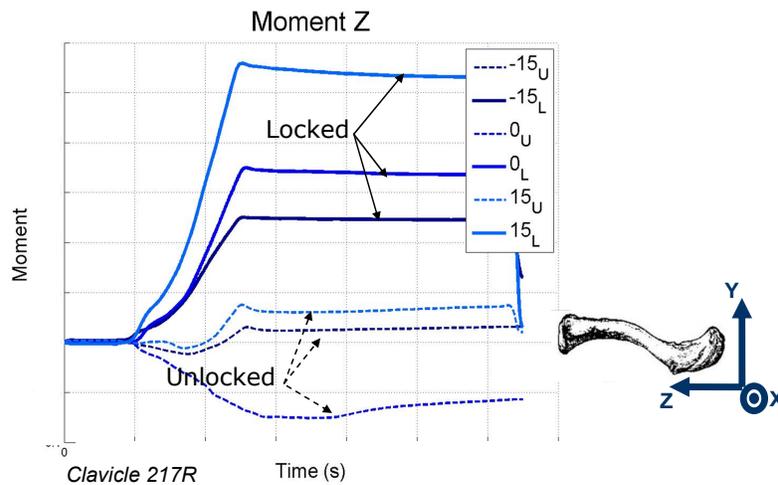

Clavicle 217R

Figure 3: Effect of locking the axial rotation on the moment around z-axis for the clavicle #217R (Solid lines: axial rotation=locked and 3 different angles (0º, +15 º,-15º); Dashed lines: axial rotation=unlocked rotation and 3 different angles (0º, +15 º,-15º)).

One of the key findings of this study is that the direction of the resulting force vector is not very sensitive to the direction of the input displacement (Figure 4). Furthermore, the force-strain relationship was not highly sensitive to the input displacement angle (Figure 5).
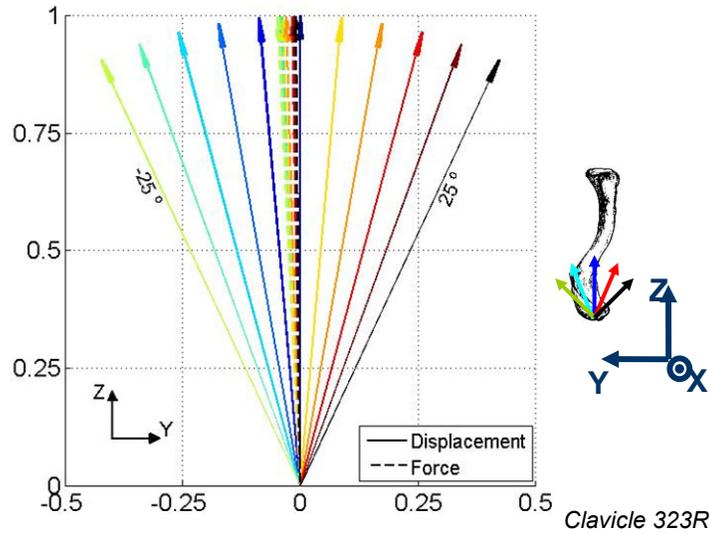

Figure 4: Force direction (dashed lines) resulting from an imposed displacement (solid lines) for different angles of the input displacement (different color for each angle).

It can also be observed on Figure 5 that the force-strain relationship is linear. The slope (i.e. the proportionality constant of this force-strain linear relationship) is different for each gage. Furthermore, the slope was found to be rate-dependent for the tested velocities.
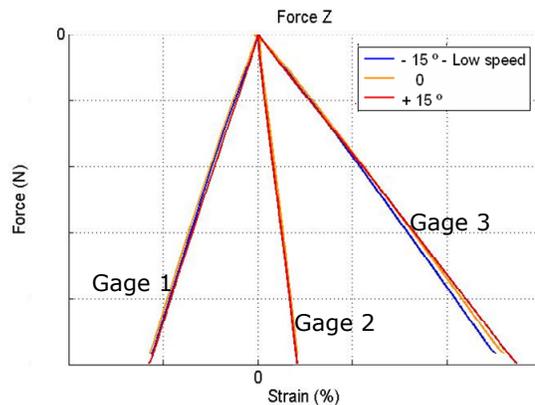
Gages around the section at the maximum of concavity (Clavicle #218R)
Figure 5: Force Z versus the strains from 3 strain gages for 3 angles, tests performed at lower velocity (different color for each angle).

DISCUSSION AND CONCLUSIONS

The present study explored a methodology to estimate the force transferred through the clavicle when submitted to compression, based on strain measurements. Quasi-static compression tests were performed on clavicles instrumented with strain gages. First, the quality of the loading fixture was assessed by ensuring that the tests were repeatable. Secondly, the chosen boundary conditions were assessed as fair as they did not engender significant moments.

It was shown that the results do not vary over a reasonable range of boundary conditions: the force-strain relationship did not depend on the z-rotation (free or locked). Furthermore, the resulting force and force-strain relationships are not dependant on the angle of the imposed displacement. Finally, it appeared that the force-strain relationship is linear at low rate.

The main limitation of this study is the rates at which the clavicles were tested, which do not correspond to rates found during car crashes. Thus, subsequent studies should assess clavicle compression at dynamic rates in order to ensure that the force-strain relationship remains linear. A second limitation is the incomplete assessment of the boundary conditions used here, and the extent to which they represent human joint behavior. Finite element simulations of clavicle compression tests would be useful in order to more fully investigate the dependence of the force-strain relationship on the boundary condition simplifications.

Thus, the results obtained in this study point out that the force-strain relationship is linear, and that the force transferred into the clavicle can likely be deduced from strain measurements if the coefficient of this relationship is known. An application of this methodology would be the possibility of finding the force transferred into the clavicle during shoulder impacts. In the case of side impact on the shoulder, the following procedure is proposed.

Intact shoulder tests would be performed with a high number, n, of gages (for example 6 gages) pasted on the clavicles, the strains ε_i $(i = 1 : n)$ would be measured as well as the displacement D and the rate $\dot{D}$ of the lateral extremity of the clavicle. Then, the clavicle would be harvested and compression tests would be performed on this isolated clavicle using the fixture described here, by applying the displacement D at the rate $\dot{D}$. From these clavicle compression tests, the coefficients K_i $(i = 1 : n)$ representing the force-strain linear relationship $Fclav_i = K_i \cdot \varepsilon clav_i$ $(i = 1 : n)$ for all n gages would be obtained. Then, using the strains measured during the shoulder tests ε_i $(i = 1 : n)$ and the coefficients K_i $(i = 1 : n)$, the force transferred into the clavicle during the shoulder tests would be found by applying $F_i = K_i \cdot \varepsilon_i$ $(i = 1 : n)$. The n equations should be redundant and result in the same force ($F_i = F$ $(i = 1 : n)$). If this is not the case, then a finite element study would be used in order to simulate the clavicle compression tests and tune the boundary conditions, until appropriate coefficients K_i $(i = 1 : n)$ would be found such that the same force estimate results from each strain measurement.

Such results would provide a better understanding of the way the shoulder is loaded during car side impacts and would definitely help the process of designing new safety devices for car lateral protection.

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DISCUSSION

PAPER: **Strain Based Methodology to Assess the Load into the Clavicle During Shoulder Lateral Impacts**

PRESENTER: ***Sonia Duprey, University of Virginia Center for Applied Biomechanics***

QUESTION: *Jerry Wong, FTSS*
That was a very good study over there. And, what do you have there? It looks like all your results show there is a linear relationship between the force and the strain. And as of now, there's no injury criteria. Do you intend to use deflection or the force from this study? Do you think there is a very positive relationship between the force and deflection so you can eliminate one? For example, I can measure the force or deflection and only either one of those can be used for injury criteria.

ANSWER: So, an injury criteria for the shoulder or--?

Q: Yes.

A: Or for the chest?

Q: For the shoulder.

A: So for the shoulder—The injury criteria you are talking about is—Usually, what is the main injury criteria for the shoulder is the deflection between the sternum and acronium because it's directly related to the length of the clavicle. So when this deflection is high, it means that you have had a clavicle injury, a clavicle fracture. So right now, I think this injury criteria is quite good. So, I don't think we should introduce the strain right now in this criteria.

Q: Right. I wasn't asking for a strain level because we're—We designed the dummy, typically for adult dummy, which are fine; and for a child dummy, it's difficult to have both measurements device over there; for example, deflection and load cells for the shoulder. So what I'm looking for is if you can either measure the force, which is preferred, or are we going to be able to eliminate the deflection as a usual criteria? That's what I'm looking for.

A: Okay. So, that's a good idea actually. We will have to investigate higher strain rate because now it's only quasi-static. But if we can find a level of strain or an interesting injury criteria including the strain, that will be very useful, actually, for injury criteria for dummy, as for children dummies. That's a very interesting point.

Q: Okay. So your view is that there is a possibility to use force as injury criteria instead of the deflection because there's an inner relationship between the two.

A: I think, yes. For the shoulder, it's maybe a possibility. Yes.

Q: Okay. Thank you.

Q: *Erik Takhounts, NHTSA*
I have a question for you, too. Right now, we have a linear relationship between the force and strain. And the way I understand it, later on you will have those strain gages and you will be able to calculate forces that act on the clavicle. Right? So, but my question is: It looked like your force levels were relatively small and strain levels. So you're not just having force relationship as the function of the strain rate that will change but also a function of the strain level, especially when it gets to the plastic region.

A: You are right: With the different strain rate, we will have different relationship. So I think when we reach the plastic level, then it's very short, then the clavicle is breaking very soon after that. So, we will just focus on this linear relationship. I mean, before.

Q: Are you going to also compress the clavicle to the moment when it breaks as well?

A: I think we will look at that, too.

Q: Okay. That's basically what I'm asking.

A: Yes.

Q: And see how the force versus strain remains linear or not.

A: Yes. Exactly. So the results here are interesting because when you are performing simulations—I mean at higher strain rates, then this relationship remains linear. But here with the experimental test, here you can see it's not totally linear anymore. So, it will be very interesting to study that. And yes, if we can go up to failure, then we will be able to study a bit what's happening, perhaps, before the plastic period.

Q: Alright. Thank you.