CHARACTERIZATION OF THE PEDIATRIC SHOULDER'S RESISTANCE TO LATERAL LOADING CONDITIONS

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ABSTRACT

Current efforts to prevent injury to children in car accidents involve the use of pediatric anthropomorphic test devices (ATDs) which are designed based on data from adult post-mortem human subjects (PMHS) and animal surrogates, rather than from data obtained directly from the pediatric population. In this study, the force-deflection characteristics of the pediatric and adult shoulder were measured directly using a combination of optical motion capture, resistive loading, and electromyography (EMG). The right shoulder of nine adult volunteers and ten pediatric volunteers was quasi-statically displaced using a hand-held force applicator in both medial and posteromedial directions. Each subject had reflective markers placed on the upper right arm, both acromions, the manubrium, and both epicondyles of the right elbow. The motions of the reflective markers were tracked using an eight-camera Vicon motion capture system. Surface EMG electrodes were applied to the latissimus dorsi, upper trapezius, anterior deltoid, posterior deltoid, biceps brachii, and pectoralis major to measure the level of muscle activity during loading. Three to five tests were performed for each loading direction and in both relaxed and tensed states. The resulting force-deflection curves were normalized and then shoulder stiffness was calculated. Shoulder stiffness in the medial direction could not be obtained since less than 2 mm of shoulder deflection was recorded in the medial loading direction prior to the data being truncated due to subject tilting. The shoulder stiffness in the posteromedial direction was found to be 3.8 N/mm for the 50th male, 2.4 N/mm for the 10 year old age group, and 3.7 N/mm for the 6 year old group in the relaxed condition. In the tensed condition, posteromedial shoulder stiffness was found to be 9.7 N/mm for the 50th male, 4.1 N/mm for the 10 year old age group, and 5.0 N/mm for the 6 year old age group. Statistical analyses were performed and it was found that adults had a significantly higher shoulder stiffness than the children. Tensed shoulder stiffness was found to be greater than relaxed shoulder stiffness for all age groups ($p < 0.001$).

INTRODUCTION

Motor vehicle crashes are a leading cause of death and disability to the pediatric population as they account for approximately 50% of pediatric trauma (Brown et. al., 2006). Even while properly restrained within a vehicle, hundreds of children are still killed or injured in motor vehicle crashes due to a lack of protection (Fildes et al., 2003; Simpson et al., 1992). This is especially true in side impacts where the risk of a child being killed was found to be much higher than frontal impacts due to the child's proximity to the side of the vehicle and the lack of available vehicle structure to absorb crash energy (Fildes et al., 2003; Franklyn et al., 2007; Simpson et al., 1992). This lack of protection is a possible explanation for why lateral impact crashes were found to represent the leading cause of injuries and fatalities to the pediatric population in motor vehicle accidents (Franklyn et al., 2007). Forty-two percent of children who were fatally injured in a motor vehicle accident were in a side impact collision (Arbogast et al., 2005). According to the Crash Injury Research Engineering Network (CIREN), children involved in side impact crashes were more likely to suffer severe injuries to the head and thorax. Of these severe injuries to pediatric crash victims, 34% were to the thorax, while approximately 43% of injuries were to the head (Brown et al., 2006).

It is important to note that the motions of the head and thorax during impact are heavily dependent on the response of the occupant's shoulder. During the event of a lateral impact, children in and out of car seats interact first with the side of the child restraint or the interior side of the vehicle, such as an intruding door. It has been observed that the shoulder is the first part of the occupant to be struck. When loaded in this manner, the shoulder deflects medially towards the thoracic cage, which results in the distribution of the initial impact load to the thorax through the spinal column, and to the head (Thollon et al., 2001). It has therefore been theorized that the skeletal components of the shoulder girdle play an important role in absorbing impact energy and reducing the energy as it is transferred to the thorax and head of the occupant.

To improve vehicle safety for children, the Q-series of child dummies was developed to cover the child population up to age 12. The Q-series was designed not only to be biomechanically advanced, but also to be used in both frontal and side impacts making it the first multi-directional series of child dummies. Unlike adult dummy development, ethics has limited the amount of child subject data available for the development of biofidelic child dummies. Therefore, the scaling of adult data is used to establish biofidelity targets for the child dummies. The scaling that was applied to the Q-series of dummies was based on the differences between adult and child subjects in terms of geometry and stiffness. The scaling factors for geometry are based on a well established set of anthropometry data for the 50th percentile male and the child anthropometry database, and the scaling factors for stiffness are based on published tissue data. Damping is not scaled due to the lack of biomechanical data, implying equal damping characteristics for children and adults (van Ratingen et al., 1997).

However, the maturity and development of a child's musculoskeletal system differ greatly from those of an adult. The bones of children are not fully ossified and are composed of a large amount of cartilaginous tissue. The muscles of children are also not as developed as those of an adult. Therefore the method of scaling geometries and stiffness to define the biofidelity response of child dummies is debatable and the overall biofidelity of child dummies is questionable.

Previous research has successfully analyzed and measured the pediatric shoulder's range of motion (Dayanidhi et al. 2005; Duff et al. 2007; Endo et al. 2004; van Andel et al. 2008; Vermeulen et al. 2002). However, these studies are not useful for modeling the response of the pediatric shoulder to impact since only the relative motion of the shoulder during everyday tasks was observed. Stiffness measurements are needed in order to develop a biofidelic shoulder in ATDs since it is important to know the amount of force required to displace the shoulder. Therefore, instead of analyzing its range of motion, a medial and posteromedial loading of the shoulder should be analyzed (Kapandji, 1982). By measuring the forces required to displace the shoulder in a manner that is similar to impact conditions, the proper shoulder stiffness can be defined for the child population, which can lead to the development of more biofidelic child ATDs.

The purpose of this study was to investigate the response of the pediatric shoulder by quasi-statically and non-injuriously analyzing its resistance to lateral loading conditions, and compare it to the response of the adult shoulder. This study was conducted in two phases. Phase one consisted of defining and validating a new method for quasi-statically measuring the shoulder's stiffness, and then quasi-static non-injurious shoulder deflections were performed on adult volunteers to define the adult shoulder's stiffness. In phase two, quasi-static shoulder stiffness testing on pediatric volunteers was performed.

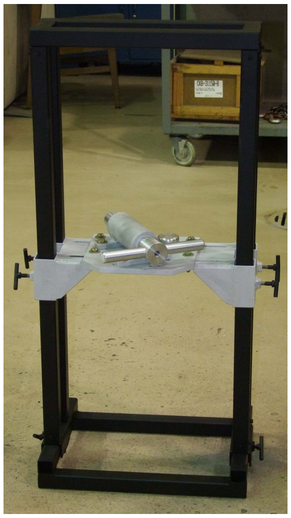
METHODS

Subjects

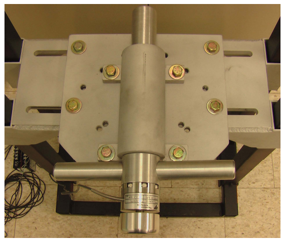
This study was reviewed and approved by The Ohio State University Institutional Review Board (IRB #2008H0202) and informed consent was obtained from all subjects. Nine adult volunteers (mean of 24 ± 3.6 years, 79 ± 10 kg) and ten pediatric volunteers (mean of 8 ± 2.3 years, 32 ± 12 kg) participated in this study. To be included, adult subjects had to be male, between the ages of 21-40 (a majority of the epiphyses have fused around the age of 21), and roughly meet the 50th percentile male requirements (78 kg, 175 cm). Children were either male or female (pediatric ATDs are representative of both male and female populations) and between 4-12 years of age. The age range of 4-12 years was chosen to correspond with the 6 year old and 10 year old ATDs. There were no height and weight requirements for pediatric subjects. The exclusion criteria for both groups were any history of injury or surgery to the shoulder, scapula, or clavicle. All male subjects were tested with their trunk bare and female subjects wore a tank-top so that the acromion was visible and to allow for non-restrictive shoulder movements.

Resistive Shoulder Loading

To measure the forces needed to displace the shoulder, a custom linear force applicator was developed utilizing a Honeywell Model 31 Mid-Range Precision Miniature Load Cell. A frame, with translational motions in the x, y, and z-directions, was designed to allow for the proper alignment of the load cell with the subject's shoulder (Figure 1). A faceplate at the center of the fixture, on top of which a load cell guide was attached, was designed to allow for a medial and posteromedial (30° anterior to medial) loading direction (Figure 1).



(a).



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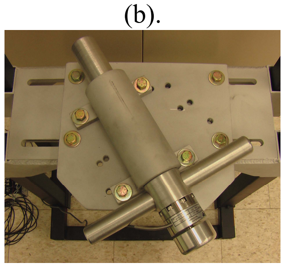


(c).

Figure 1. (a). Frame designed for the proper alignment of the shoulder force applicator with the subjects' shoulders; (b). Load cell attached to the tip of the force applicator and positioned in a medial loading direction; (c). Force applicator positioned in a posteromedial loading direction.

To measure shoulder girdle deflection and thoracic motion, an 8-camera, 100 Hz Vicon motion analysis system (Vicon Motion Systems, Oxford, UK) was used. Reflective markers were placed on the skin using double-sided adhesive tape over the subject's acromion process of both scapulas, manubrium of the sternum, lateral and medial epicondyles of the right humerus, and around the mid-shaft of the humerus as an 8-marker cluster with a 2x1x2x1x2 configuration (Figure 2). In addition, reflective markers were placed on the load cell guide and the bench on which the subjects were seated.

To measure muscle activity during the tests, surface electrodes were applied to the superficial muscles that play an important role in the movement and stabilization of the shoulder. The muscles documented and analyzed were the latissumus dorsi, upper aspect of the trapezius, anterior and posterior portions of the deltoid, biceps brachii, and pectoralis major of the displaced shoulder.

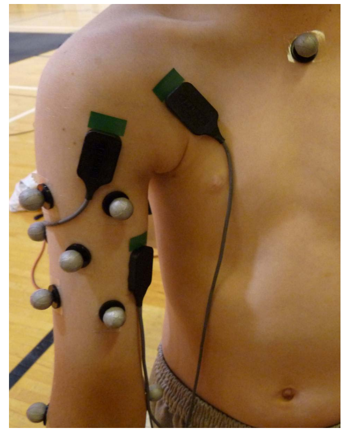


Figure 2. Placement of reflective markers and surface electrodes.

The subject bench and shoulder apparatus were placed at the center of the 8-camera Vicon optical motion capture setup. Each subject was seated on the right edge of the bench with the right side of the seatback along the subject's spine, allowing for a free range of motion of the shoulder. With the subject in position, a hip brace was applied to the subject's left hip and clamped onto the bench to prevent any translational motion of the subject's pelvis during the various loading conditions (Figure 3).

Figure 3. Image of an adult test setup. The custom linear force applicator and frame are seen to the subject's right side. A special bench with Teflon backing was used to allow for a free range of motion of the shoulder. A hip brace was placed to the subject's left to prevent total body sliding during testing.

Prior to the positioning of the load cell, the maximum voluntary contractions (MVCs) of the muscles were recorded. With the aid of a researcher, the subject's right arm was placed at prescribed positions and then the subject was told to move his or her arm in various directions with as much force as possible while the researcher provided resistance such that each relevant muscle was maximally contracted. The subject maintained each maximal contraction for five seconds, was told to relax for five seconds, and then told to repeat the movement and maintain for another five seconds while recording the signal. Upon completion of the recording of the MVCs, the shoulder loading apparatus was placed next to the subject. The height of the load cell was adjusted and centered at the lateral portion of the subject's deltoid muscle covering the glenoid fossa of the scapula. For each test, a researcher would manually push the load applicator and slowly displace the subject's shoulder. As the subject began to tilt, the application of the force was terminated and the load applicator was retracted back to its starting position. Three to five tests were performed with the subject's muscles relaxed, and three tests were performed with the subject's muscles tensed. The sequence of relaxed tests followed by tensed tests was performed for both medial and posteromedial loading directions.

Data Reduction and Analysis

Marker data acquisition was performed at 100 Hz and processed using Vicon Nexus software. Forces from the load cell were acquired at 1000 Hz and filtered using a low-pass butterworth filter at 100 Hz. EMG signals were acquired at 1000 Hz, rectified, filtered using a bandpass filter between 10 and 400 Hz, and then filtered using a low-pass filter at 25 Hz for analysis. The deflection of the shoulder was calculated using the Vicon marker data as the change in distance between the acromion of the shoulder being displaced and both the manubrium (half-girdle deflection) and non-displaced acromion (full-girdle deflection), and was plotted against the applied load. Half-girdle and full-girdle shoulder deflections were used interchangeably since initial analysis of the two measurements found them to be near identical. Therefore, for each test the deflection measure that produced the largest linear region in the force-deflection curve was chosen, since ultimately the linear region of the curve would be used to calculate stiffness values. Additionally, if either the manubrium or opposite acromion were lost during motion tracking, the deflection measure that was used for the test was the one that had the available markers.

The resulting force-deflection curves were to be used to determine shoulder stiffness for each subject, loading direction, and relaxed or tensed test condition. In all cases, the force-deflection curves displayed a relatively linear response until the force became large enough to cause the subject to start tilting away from the loading. Once the subject began tilting away the force-deflection response became nonlinear and unpredictable, thus the data was effectively meaningless at that point. In some instances the subject's response to a force high enough to cause them to tilt away was to avoid it by leaning into the load, also resulting in nonlinear and meaningless force-deflection responses. Therefore in order to calculate the shoulder stiffness for each test, a linear portion of each curve had to be defined. First, data was truncated at the point where four degrees of subject tilt was observed, where tilt was defined as the change in angle between a line going through both acromions and the horizontal plane (Bolte et al., 2000; 2003) Next, the linear portion of the curve was determined by evaluating the central portion of the force-deflection curve (20-80%) and finding the range in which the slope of each point remained within one standard deviation of the average of the slopes of the previous points (Margulies & Thibault, 2000). Once the slope exceeded one standard deviation (i.e., became nonlinear) the data was truncated at that point and the data that remained exhibited a relatively linear force-deflection response that could be used for calculating stiffness.

Example force-deflection curves for the medial and posteromedial loading conditions of adult subject 7 are shown in Figures 4 and 5. The curves show the repeatability and linearity of the test trials as the curves follow similar, linear trends. Note that less than 2 mm of shoulder deflection was recorded in the medial loading direction prior to the data being truncated due to subject tilting. This was found to be the case for all pediatric and adult subjects in the medial direction and indicates that the clavicle is stiff enough that no appreciable shoulder deflection can be achieved in this quasi-static manner before the subject begins to tilt. Since the average human skin thickness ranges from 0.5 mm (eyelids) to 4 mm (soles of hands and feet), the 2 mm of deflection observed in these tests is on the order of what would be required simply to compress the skin on the shoulder. In addition, the resolution of accuracy for the Vicon motion capture system is on the order of 0.1 mm which means a minimum of 5% error would immediately be introduced into the deflection measurements. For these reasons, only the posteromedial stiffness is reported and discussed in this paper since it is the opinion of the authors that 2 mm of shoulder deflection is not suitable for calculating shoulder stiffness. Similarly, data from any test trials in the posteromedial loading direction in which the shoulder deflection did not exceed 2 mm was also excluded from analysis. Future testing will incorporate a load wall positioned at the subject's non-loaded shoulder to prevent immediate tilting and allow for more shoulder compression. Preliminary pilot studies have demonstrated that much more shoulder deflection can in fact be achieved in the medial loading direction using an opposing load wall, and will be presented in a future publication.

In order to calculate stiffness for each subject the repeated trials in each test condition were reduced to one representative mean force-deflection curve for each subject and test condition. However, since the quasi-static load for each test was applied manually, the loading rate was not controlled resulting in different loading rates for each test. Therefore, the typical calculation of the mean and standard deviation of a set of curves using the time-histories is invalid since forces and deflections from each trial used to calculate the mean were reached at different times. Since a representative mean curve was still desired for each subject and test condition, the repeated trials within each group were interpolated onto common values of deflection and then a mean curve was calculated in "force-deflection space" instead of using the time-histories. It should be noted that since the data was interpolated on common levels of deflection, and no extrapolation of the data was performed, the mean curve could only be calculated up to the smallest value of maximum shoulder deflection of each trial since force data was not available at any further values of deflection for that trial. Once the force-deflection curve was reduced in this manner to a single mean curve for each subject and test condition, a linear fit could be obtained, taking the slope to be the shoulder's stiffness.

Normalization

All adult data were normalized to the anthropometry of a 50th percentile male, data from pediatric subjects age 8 to 12 were normalized to the anthropometry of the 10 year old ATD, and data from pediatric subjects age 4 to 7 were normalized to the anthropometry of the 6 year old ATD.

The underlying basis of the normalization procedure was a spring-mass model first introduced by Mertz (1984) which incorporates a mass ratio and a stiffness ratio. In Mertz (1984), the mass ratio was comprised of effective mass values calculated from the subject response data using an impulse-momentum analysis. The denominator of the ratio was the effective mass calculated for each individual subject. The numerator was determined by calculating the percentage of each subject's effective mass to their total body mass, averaging the percentage across subjects, and multiplying by the total body mass of the population to which the data was to be normalized (e.g., 76 kg for 50th percentile male). The stiffness ratio was simply a ratio of characteristic lengths (e.g., chest depth) where the denominator was the characteristic length of the subject and the numerator was the characteristic length of the population to which the data was to be normalized. Moorhouse (2011; 2008) took this methodology a step further by also incorporating the response data into the determination of the stiffness ratio. Using a procedure analogous to the effective mass ratio described above, the denominator was determined by calculating the effective stiffness of each subject from the response data, and the numerator was determined by calculating the percentage of each subject's effective stiffness to a characteristic length of the subject, averaging the percentage across subjects, and multiplying by the characteristic length of the population to which the data was to be normalized.

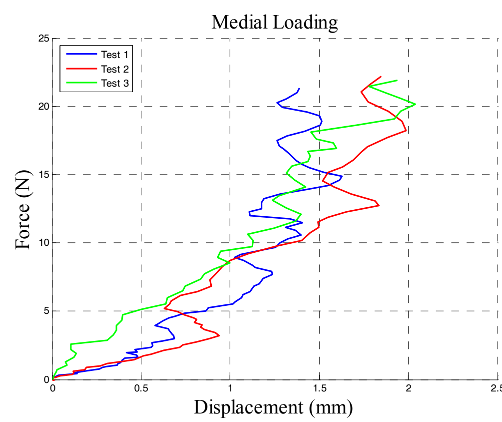


Figure 4. Force-deflection curves for adult subject 7 in the medial loading condition.

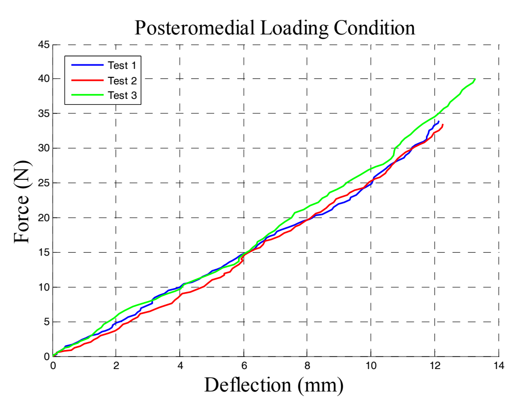


Figure 5. Force-deflection curves for adult subject 7 in the posteromedial loading condition.

For the data from the current study, the impulse-momentum procedure used to calculate the effective mass of each subject is not valid since the quasi-static loading rate does not represent impulse loading. Therefore it was decided to calculate the mass ratio (λ_m) using the total body mass of the subject for the denominator and the total body mass of the population to which the data was to be normalized for the numerator (Equation 1). To determine the stiffness ratio in this study, first an effective stiffness (k_{eff}) for each subject and test condition was calculated using Equation 2. Then a percent stiffness ratio (%Stiff) was calculated by dividing the subject's effective stiffness by their shoulder breadth, where the shoulder breadth was deemed the most appropriate characteristic length for this study. Within each test condition the values for %Stiff were averaged across subjects (Avg %Stiff), and finally the stiffness ratio (λ_k) was calculated using Equation 3.

$$\lambda_m = \frac{Mass_{50th,10yo,6yo}}{Subject\ Mass} \quad (1).$$

$$k_{eff} = \frac{2\int F dx}{x_{max}^2} \quad (2).$$

$$\lambda_k = \frac{Avg\ \%Stiff \times Shoulder\ Breadth_{50th,10yo,6yo}}{k_{eff}} \quad (3).$$

Normalizing factors for force (λ_F) and deflection (λ_D) were then calculated from the resulting mass and stiffness scaling ratios (Equations 4 and 5), the normalized force and deflection were cross-plotted, and the normalized shoulder stiffness determined via the slope of a linear fit.

$$\lambda_F = \sqrt{\lambda_m \times \lambda_k} \quad (4).$$

$$\lambda_D = \sqrt{\lambda_m / \lambda_k} \quad (5).$$

RESULTS

Important subject information taken from anthropometric measurement sheets are listed in Table 1. Subjects were divided into three age groups corresponding to the 50th percentile male, 10 year old ATD, and 6 year old ATD, as described in the normalization section above.

Force-deflection plots of the repeated trials for each subject and test condition which were used to generate the non-normalized mean force-deflection curves are provided in Appendix A. Appendix B contains the six sets of the non-normalized mean force-deflection curves for each age group and test condition.

After normalizing the data, biomechanical targets were created for each age group and test condition to represent the force-deflection response of the adult, 10 year old, and 6 year old shoulder to quasi-static posteromedial loading in relaxed and tensed conditions (six total biomechanical targets). For the same reasons described in the methods, mean and standard deviation curves for each age group and test condition could not be calculated using the time-histories. Instead, data from all subjects within a test condition were first interpolated onto common values of deflection and a mean curve and force standard deviations were calculated. Next the data was interpolated onto common values of force so that the deflection standard deviations could be calculated. The resulting deflection standard deviations were then interpolated onto the mean curve so that the force standard deviations and deflection standard deviations occurred at common points. Finally, ellipse targets were developed using the force and deflection standard deviations by calculating an ellipse at each point along the mean force-deflection curve as previously described in Shaw (2006).

Relaxed and tensed biofidelity targets for each of the age groups are plotted in Figures 6 and 7, respectively, along with the normalized mean force-deflection curve for each individual subject.

A summary of normalized shoulder stiffness for the 50th male, 10 year old, and 6 year old in both relaxed and tensed conditions is shown in Table 2. Statistical analysis using a two sample t-test assuming unequal variance was performed on both the relaxed and tensed stiffness data. The resulting p-values are tabulated in Table 3 and show that all three age groups demonstrate statistically significant differences in shoulder stiffness for both relaxed and tensed conditions (adult >> 6YO >> 10YO). In addition, tensed shoulder stiffness was found to be greater than the relaxed stiffness in all three age groups ($p < 0.001$).

Table 1.
Subject age and anthropometry data

	Subject #	Age	Gender	Mass (kg)	Seated Height (cm)	Shoulder Breadth (cm)
Adult	**1**	23	M	82	84	43
	2	24	M	77	91	48
	3	23	M	80	91	43
	4	23	M	73	94	39
	5	32	M	79	93	40
	6	22	M	70	95	44
	7	25	M	73	91	39
	8	20	M	74	90	40
	9	28	M	102	91	39
	Average	24 ± 4		79 ± 10	91 ± 3	42 ± 3
10YO	**P1**	10	M	50	74	34
	P4	8	M	39	73	36
	P7	9	M	34	74	29
	P8	11	M	43	82	35
	P10	8	F	25	68	29
	P11	10	F	43	80	39
	Average	9 ± 12		39 ± 8	75 ± 5	33 ± 4
6YO	**P3**	7	M	26	68	28
	P5	5	F	16	56	22
	P9	4	F	18	70	20
	P12	6	F	23	66	28
	Average	6 ± 1		21 ± 5	65 ± 6	24 ± 4

Table 2.
Normalized posteromedial shoulder stiffness (N/mm) for the 50th male, 10 year old, and 6 year old age groups

	Relaxed	Tensed
50th Male	3.84	9.69
10YO	2.44	4.11
6YO	3.67	4.98

Table 3.
Statistical significance between normalized shoulder stiffness for the three age groups (two sample t-test assuming unequal variance)

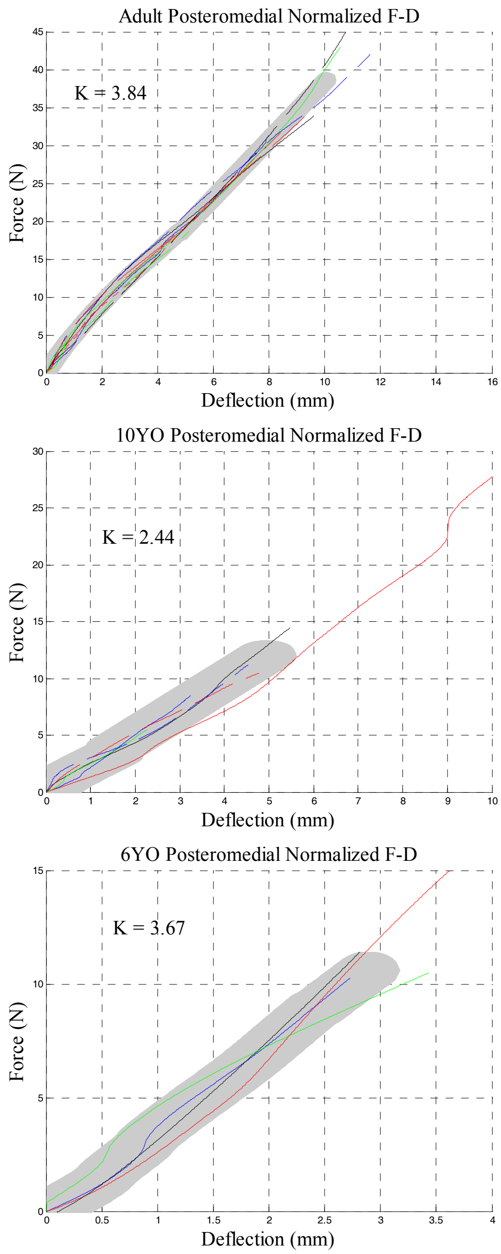
p-Values		
Age Group Comparison	**Relaxed**	**Tensed**
Between 10YO and 6YO	< 0.001	< 0.05
Between 10YO and Adult	< 0.001	< 0.001
Between 6YO and Adult	< 0.05	< 0.001

DISCUSSION

This study investigated the response of the shoulder to lateral loading by quasi-statically and non-injuriously analyzing its resistance to lateral loading conditions. A total of 9 adult and 10 pediatric volunteers were tested, and the stiffness of the shoulder in a posteromedial loading direction in both relaxed and tensed conditions was obtained.

A cursory examination of Tables 2 and 3 reveals that for both relaxed and tensed conditions the shoulder stiffness in the posteromedial direction of all three age groups are statistically different from one another (i.e., adult >> 6YO >> 10YO), and that for all three age groups the tensed shoulder stiffness is greater than the relaxed shoulder stiffness. However, examination of Table 4 which lists the individual stiffness values (both non-normalized and normalized) for each age group and test condition reveals that the normalization procedure drastically reduces the variance in stiffness within each age group, potentially resulting in inflated statistical significance between age groups.

If the non-normalized stiffness for each age group is evaluated for statistical significance (Table 5), it can be observed that in the relaxed condition there is still



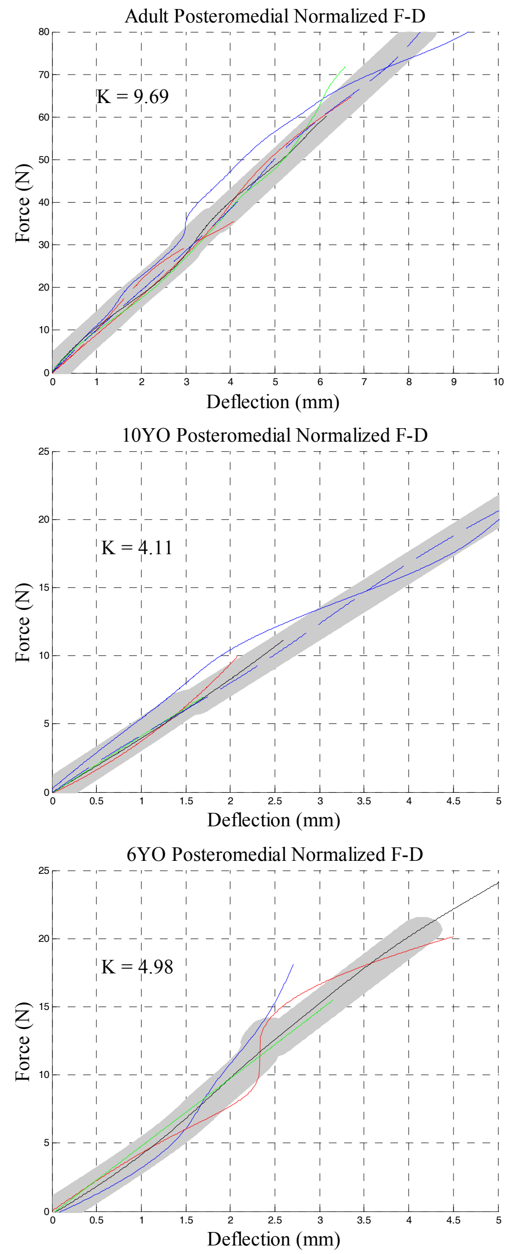


Figure 6. Normalized force-deflection curves and biofidelity targets (grey) for the 50th percentile adult male, 10 year old, and 6 year old in the relaxed posteromedial loading condition. The targets were created by forming one standard deviation ellipses around the mean force-deflection response.

Figure 7. Normalized force-deflection curves and biofidelity targets (grey) for the 50th percentile adult male, 10 year old, and 6 year old in the tensed posteromedial loading condition. The targets were created by forming one standard deviation ellipses around the mean force-deflection response.

Table 4.
Shoulder stiffness in the posteromedial loading direction in both relaxed and tensed conditions. Adult subjects 3, 4 and pediatric subject 1 were not included in the tensed condition due to shoulder deflections that were less than 2 mm.

	Subject #	Relaxed		Tensed	
		Non-Normalized	Normalized	Non-Normalized	Normalized
Adult	**1**	6.10	3.88	9.58	9.34
	2	5.50	3.84	6.23	9.86
	3	3.14	3.78	--*	--*
	4	3.35	3.88	--*	--*
	5	3.22	3.79	10.07	9.74
	6	1.37	3.79	7.61	9.91
	7	2.36	3.96	15.58	9.74
	8	2.63	3.80	4.79	9.57
	9	4.67	3.86	--*	--*
	Mean	**3.59**	**3.84**	**8.98**	**9.69**
	Std. Dev.	**1.45**	**0.06**	**3.47**	**0.19**
10YO	**P1**	2.70	2.50	--*	--*
	P4	2.15	2.57	3.35	3.97
	P7	1.98	2.47	3.73	4.26
	P8	3.28	2.41	4.99	4.13
	P10	2.65	2.37	3.21	4.10
	P11	1.91	2.35	8.98	4.07
	Mean	**2.45**	**2.44**	**4.85**	**4.11**
	Std. Dev.	**0.48**	**0.08**	**2.16**	**0.10**
6YO	**P3**	2.16	3.69	3.41	5.30
	P5	2.35	3.80	7.34	4.90
	P9	3.97	3.76	4.35	4.85
	P12	3.40	3.42	1.65	4.87
	Mean	**2.97**	**3.67**	**4.19**	**4.98**
	Std. Dev.	**0.74**	**0.15**	**2.06**	**0.19**

a significant difference between adults and the 10 year old age group, but not between adults and the 6 year old age group, or between the 6 year olds and 10 year olds. However, in the tensed condition there is a significant difference between adults and both child age groups, but no significant difference between the two child populations. As with the normalized shoulder stiffness, all three age groups demonstrate a significantly higher shoulder stiffness in the tensed condition than in the relaxed condition ($p < 0.001$).

Table 5.
Statistical significance between non-normalized shoulder stiffness for the three age groups (two sample t-test assuming unequal variance)

p-Values		
Age Group Comparison	**Relaxed**	**Tensed**
Between 10YO and 6YO	0.16	0.35
Between 10YO and Adult	< 0.05	< 0.05
Between 6YO and Adult	0.19	< 0.05

Despite the difference in potential conclusions drawn from the normalized stiffness values versus the non-normalized stiffness values, both analyses produce clear evidence that there is a difference in shoulder stiffness between children and adults. The specific details of the difference likely lie somewhere between, and further testing with a much larger sample from each age group should help elucidate those details.

In addition, it is important that the shoulder stiffness in the medial loading direction is determined for each age group to supplement the results from the posteromedial loading direction, as it is expected that much more variation between children and adults would be seen in the medial direction. Whereas in the posteromedial direction where the stiffness of soft tissue contributing to the anterior-posterior resistance may differ to some degree between children and adults, the resistance to medial loading is primarily provided by the clavicle so bone maturity may play a large role in the response. In adults, the clavicle and other bony structures of the shoulder are fully ossified. However, in children, especially those under the age of 12, a larger portion of bones are still cartilaginous. The presence of cartilage can lead to a more compliant shoulder and result in a lower stiffness in the younger age groups. The intent of this study was to evaluate both loading directions but the combination of the quasi-static loading and the stiffness of the clavicle caused there to be no appreciable shoulder deflection (< 2 mm) before the subject began to tilt away from the loading. This could be avoided by using an opposing load wall on the non-loaded shoulder and investigation of the force-deflection response of the shoulder in the medial loading direction using an opposing load wall is currently underway and will be presented in a future publication.

The importance of the medial loading direction can be seen to some extent in the data obtained from this study in the posteromedial loading direction if the resultant force and deflection is broken down into its medial-lateral (y-direction) and anterior-posterior (x-direction) components. As expected, all three age groups demonstrate that the shoulder is much less stiff in the x-direction due to the lack of bony structures to impede the motion of the joint posteriorly. When moving in this direction, the shoulder pivots at the sternoclavicular joint and there are no bony structures that directly inhibit the shoulder's motion, thus soft tissue, rather than hard bony tissue, contributes more to the stiffness. In contrast, the shoulder has higher stiffness in the y-direction because the clavicle serves as a strut to hinder the medial motion of the shoulder, thus hard tissue contributes more to the stiffness.

The shoulder stiffness comparisons between age groups in this study yielded two unexpected findings that warrant some further discussion. First, for the normalized stiffness in both the relaxed and tensed conditions the shoulder stiffness of the 6 year old is significantly larger than the 10 year old. Due to the fact that this result did not hold for the non-normalized stiffness values, this may be an artifact of variance reduction in the normalization process which inflates the statistical significance between stiffness values from each population. The other unexpected finding was that both the non-normalized and normalized shoulder stiffness of the six year old was not significantly different from the adult in the relaxed condition, although it was significantly lower (as expected) in the tensed condition. Although these phenomena should be better understood after future studies involving a larger sample of children (only four 6YO subjects and six 10YO subjects in this study), and when stiffness data for medial loading is available, they still may be worth some consideration. It is possible that these results could be due to a lack of muscle control in younger subjects. Even when relaxed, younger subjects in the 6 year old group may involuntarily activate their muscles as a shoulder protection mechanism, which can lead to higher stiffness values than the subjects in the 10 year old group. This could also explain why in the relaxed condition the shoulder stiffness of the 6 year old group was similar to adults since their relaxed stiffness was higher due to muscle activation.

Evidence for this can be found upon close examination of the EMG data. Appendix C contains a plot of the EMG signals for a subject from each age group in both relaxed and tensed test conditions, along with a table showing the maximum %MVC obtained during each test. It can be seen that the adults demonstrate a much higher difference in %MVC between the relaxed and tensed conditions than either of the child age groups. Also, as the age of the group decreases from adult down to six year old, the %MVC in the relaxed condition increases (3%, 9%, 17%, respectively) whereas in the tensed condition is remains relatively consistent (22%, 16%, 20%).

One definitive conclusion from this study is that for all three age groups, and for both the normalized and non-normalized stiffness values, that the stiffness of the shoulder is greater with muscle tensing than when relaxed. This result is not surprising for this quasi-static loading condition because tensing of the muscles should result in stabilization of the shoulder joint and less movement for a given applied force. However, as the applied force becomes very large (i.e., well above the increase in stabilizing force of the tensed muscles) with a much higher severity of loading as seen in a crash-scenario, the relative effect of tensing on the resistance to loading would be expected to decrease. Therefore, the relevance of this result to a crash-scenario where an occupant's muscles may be relaxed or tensed depending on if they are aware of an oncoming accident is unknown. This could potentially be investigated by applying crash-level loading to the shoulder of PMHS in both tensed and relaxed conditions, where the tensed condition could be simulated using muscle stimulation to cause the muscles to contract upon loading.

Although the statistical significance of the normalized stiffness values should be taken with caution, it should be pointed out that the normalized data is very important for creating biomechanical targets for assessing the biofidelity of existing child ATD shoulders and for designing new ATD shoulders based on measured differences between children and adults. The extreme amount of variation seen in the non-normalized force-deflection response of the shoulder (Appendix B), particularly in the adults, must be reduced so that an ATD is held to higher standard when trying to match a biofidelic mean response.

LIMITATIONS

There are several factors to consider when interpreting the results of this study. First, only a limited amount of subjects (9 adults and 10 children) were tested in this study. More volunteers will need to be tested in the future in order to increase the statistical significance of the conclusions. Since completion of this study, approximately twenty additional pediatric subjects have been recruited, and testing is currently ongoing to obtain shoulder stiffness data to supplement the current study. This ongoing study will also obtain meaningful data in the medial loading direction by utilizing an opposing wall on the non-loaded shoulder so that differences in the response of the clavicle between children and adults can be taken into account

Also, the test procedure and analysis itself proved to be very challenging. Quasi-static shoulder deflections such as were performed in this study had not previously been conducted. The shoulder joint is complicated to study due to the fact that it is a floating joint and relies mostly on muscles for stabilization. Since muscle mass and muscle tone vary greatly between individuals due to their body shapes and the types of activities they take part in, the motions of the shoulder can vary greatly. Also, performing tests on volunteers provided additional challenges and introduced several factors that were hard to control in the study, especially with the younger age groups. Even though steps were taken to try and control the posture of each individual, no two volunteers sat on the bench in precisely the same manner. Subjects' shoulders were hunched or arched back; heads were leaning forward or backward; backs were straight, arched, or hunched; and the younger age groups would sometimes move around between tests and even during some tests. Another factor that was difficult to control was the amount of voluntary and involuntary muscle activation during both the relaxed and tensed testing conditions. Even when relaxed, an individual may involuntary activate some of their muscles, especially with the younger age groups who do not have full control over their muscles and may reflexively guard their shoulders. It is even very difficult for individuals to activate the same muscles, and to the same degree of activation, from test to test. Furthermore, the muscles naturally activated during "tensing" can vary greatly between individuals. All of these variables undoubtedly introduced some test-to-test variation and may have affected the results.

Despite these limitations, these results are still important in trying to understand and characterize the difference in the pediatric shoulder's resistance to various loading conditions with respect to the resistance of the adult shoulder.

CONCLUSION

Based upon the research presented in this paper, the following observations were made:

- Despite the difficulty in controlling the test conditions using volunteer subjects, the method presented for quasi-statically displacing the shoulders of adult and pediatric volunteers was repeatable as demonstrated by the similarity of repeated trials for each subject and test condition.
- The shoulder stiffness of the 50^{th} percentile adult male is significantly larger than the shoulder stiffness of children.
- Relative shoulder deflection measured from acromion-to-sternum is very similar to measured shoulder deflection from acromion-to-acromion.
- The tensing of the shoulder muscles causes an increase in shoulder stiffness
- Improvements to the test procedure were identified in this study, particularly the use of an opposing load wall on the non-loaded shoulder to prevent subject tilting. This should allow for appreciable shoulder deflection to be measured in the medial loading direction in future testing.
- The statistical power of the results could be improved by obtaining more pediatric subjects, and a study involving approximately 20 more pediatric subjects using an opposing load wall is already underway.
- Despite the limitations, the results are a good start to understanding the differences in shoulder stiffness between pediatric and adult subjects, which can hopefully lead to improved methods for developing pediatric ATDs.

ACKNOWLEDGEMENTS

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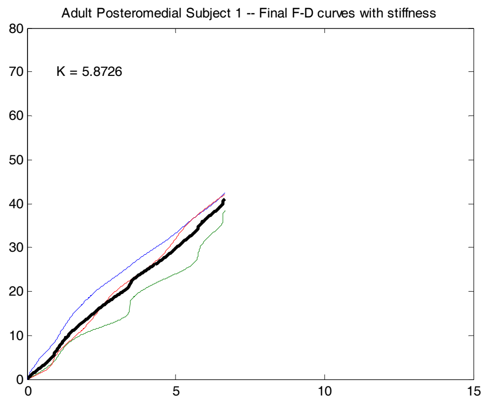
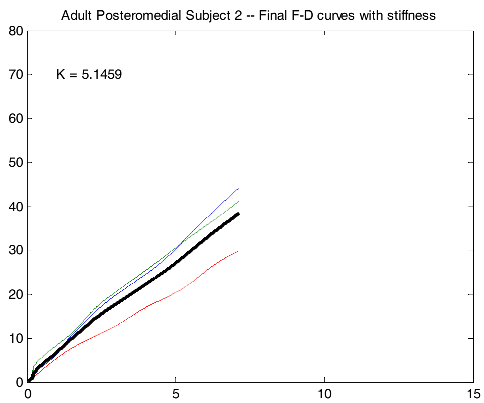
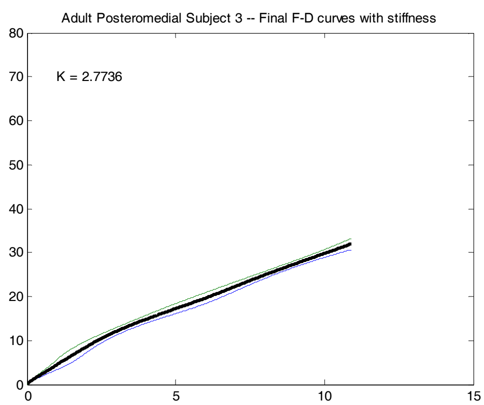
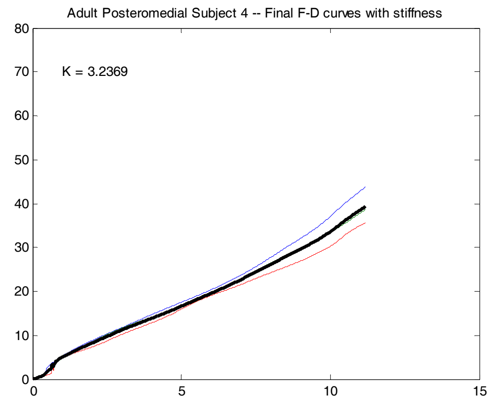
APPENDIX A

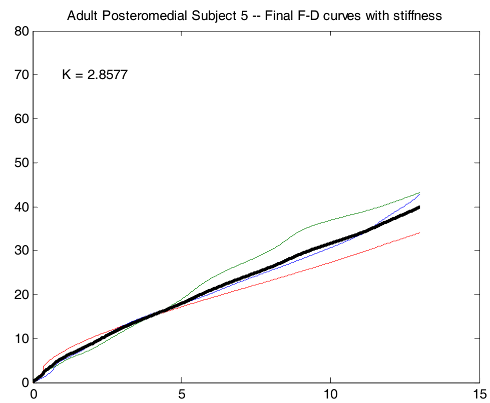
Relaxed Adult Posteromedial F-D Curves

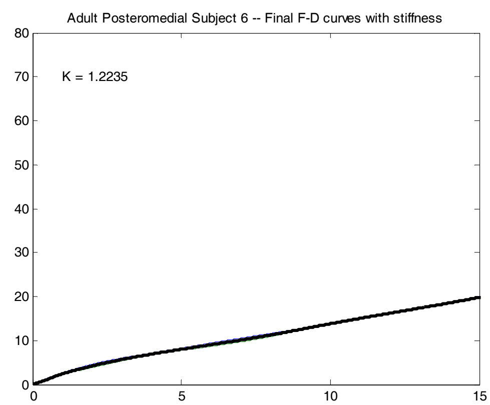


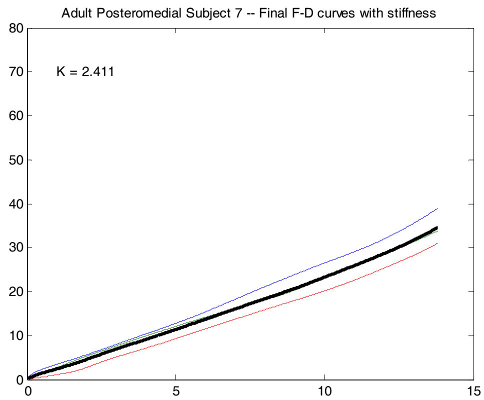











Adult Posteromedial Subject 7 -- Final F-D curves with stiffness
K = 2.411
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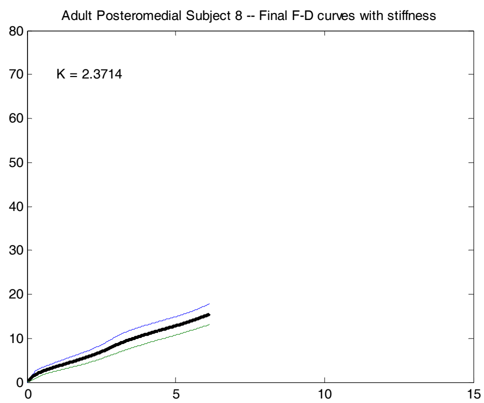
70
60
50
40
30
20
10
0
0
5
10
15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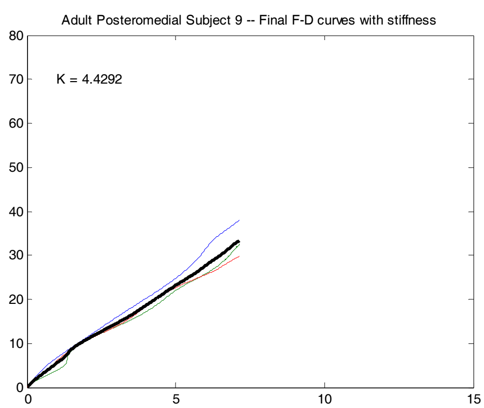
20
10
0
0
5
10
15


Adult Posteromedial Subject 9 -- Final F-D curves with stiffness
K = 4.4292
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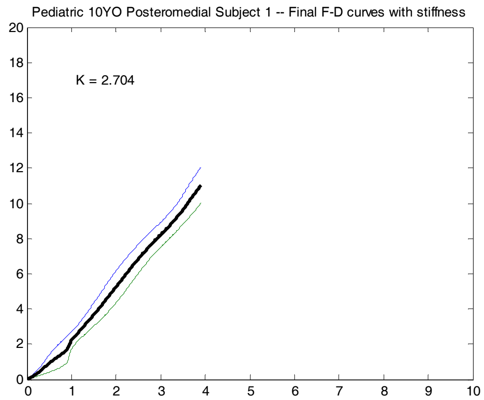
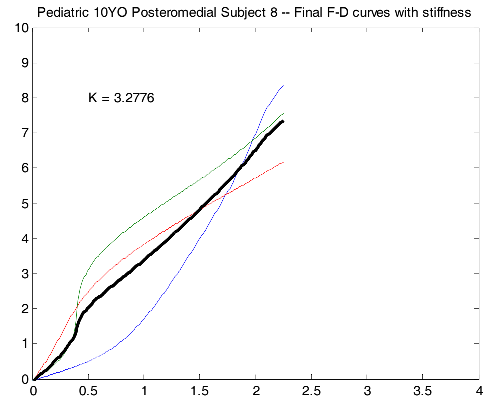
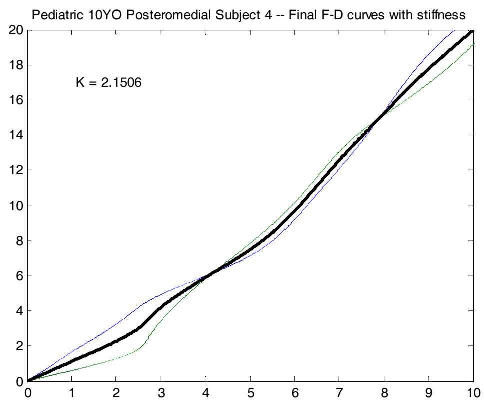
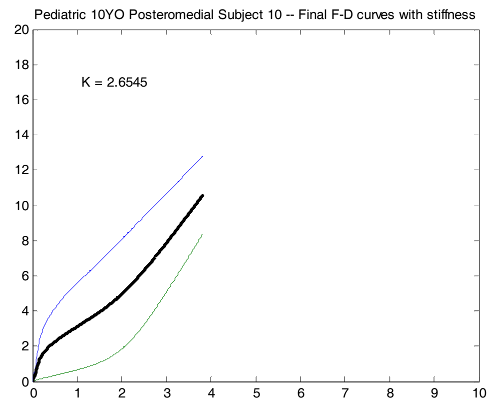
60
50
40
30
20
10
0
0
5
10
15

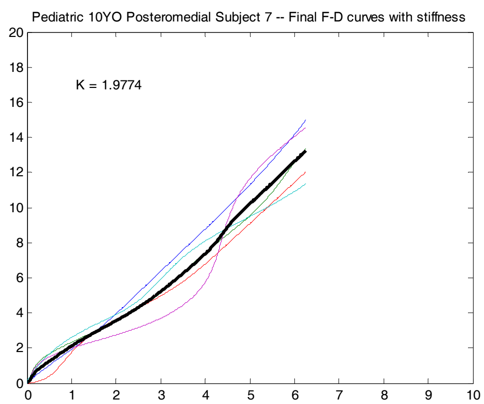
Relaxed 10YO Posteromedial F-D Curves

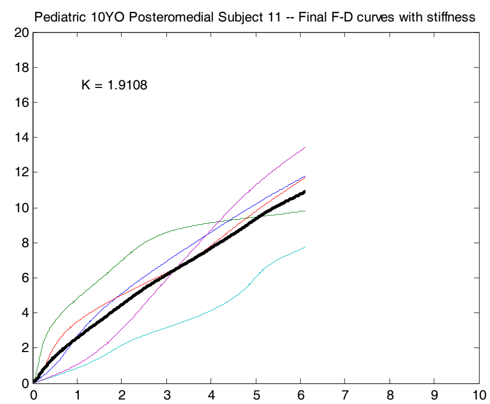




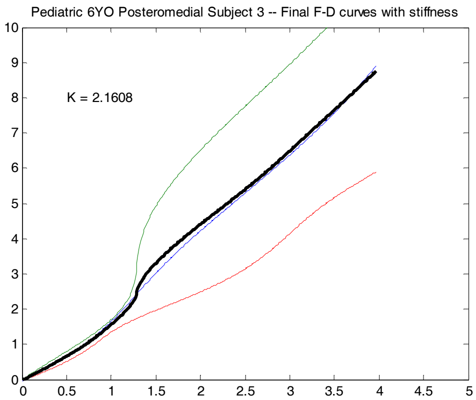


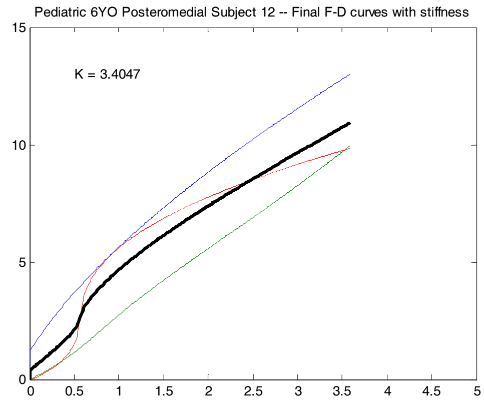


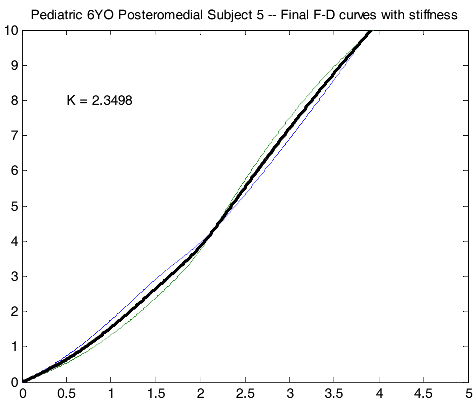


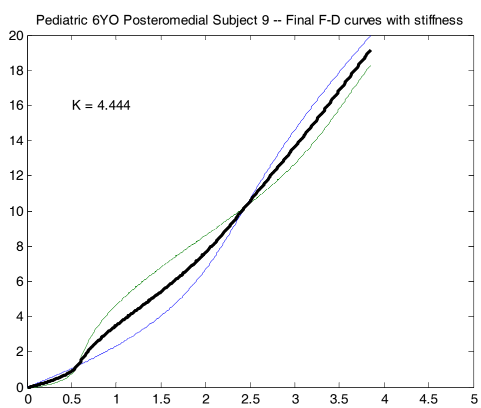


Relaxed 6YO Posteromedial F-D Curves

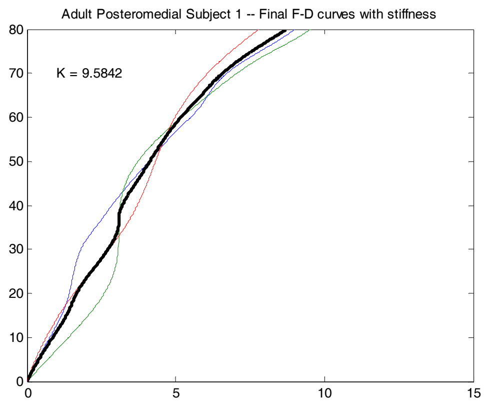


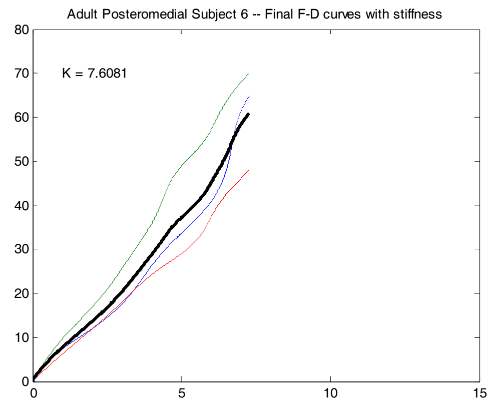


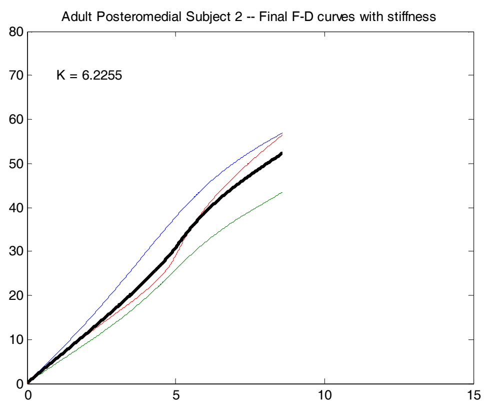


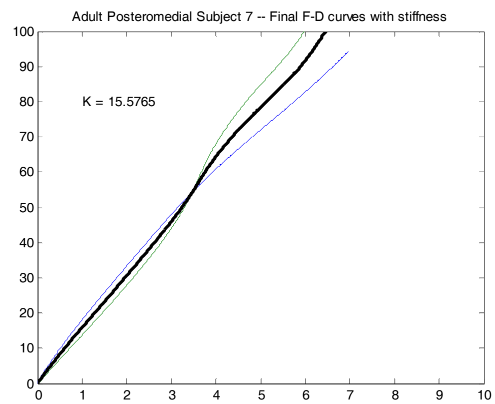


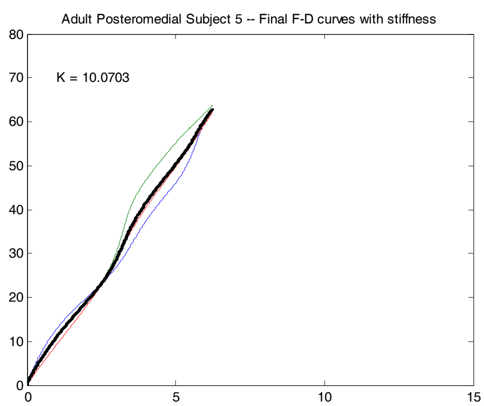
Tensed Adult Posteromedial F-D Curves

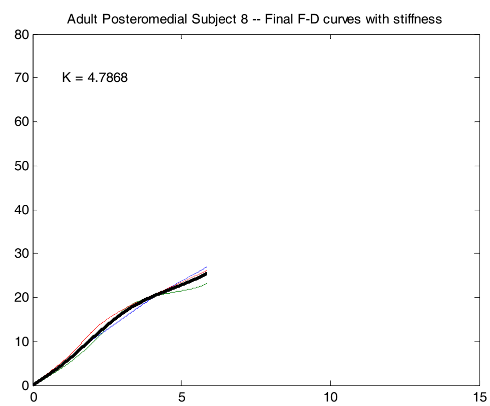




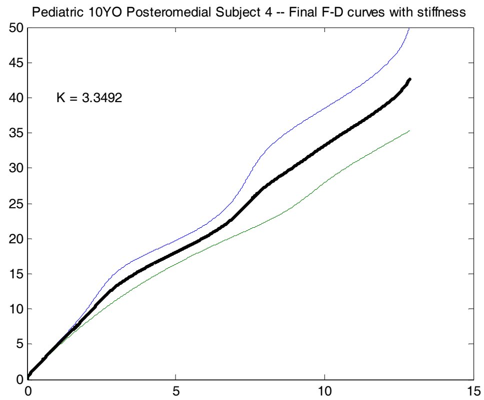


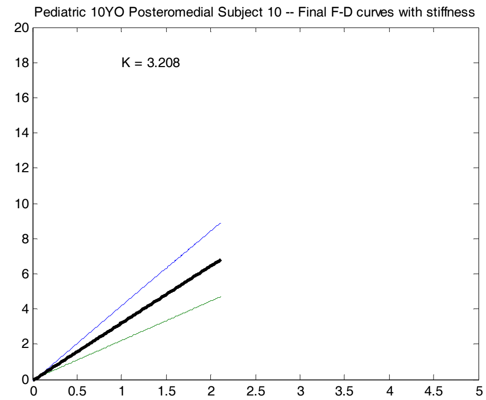


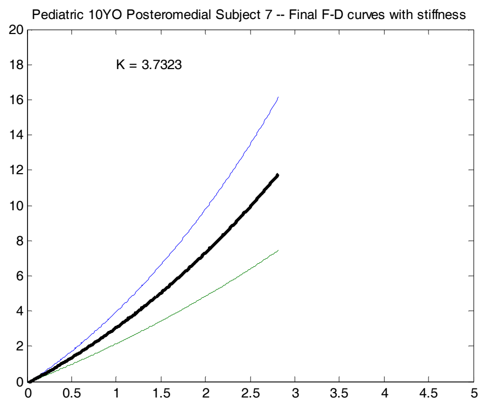


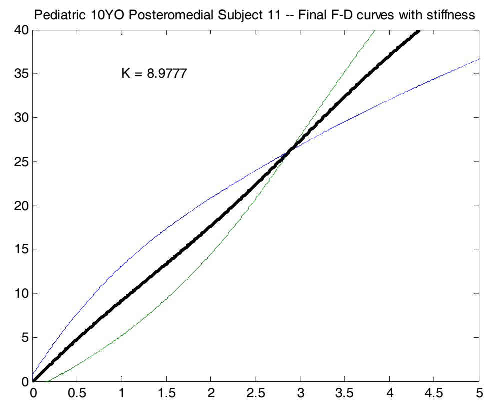


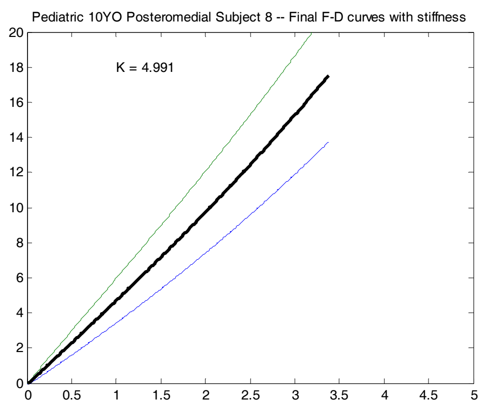
Tensed 10YO Posteromedial F-D Curves



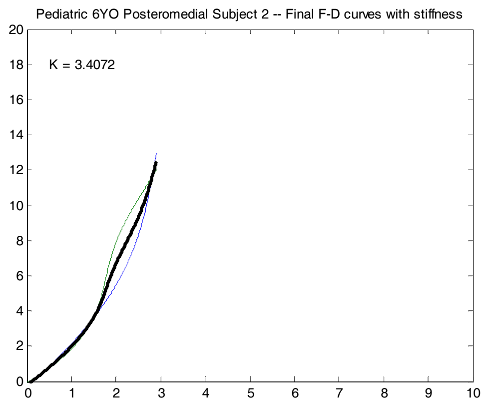


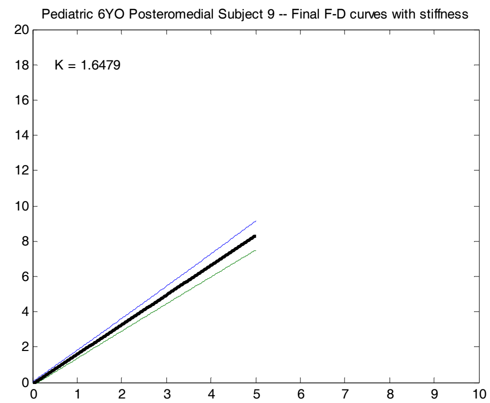


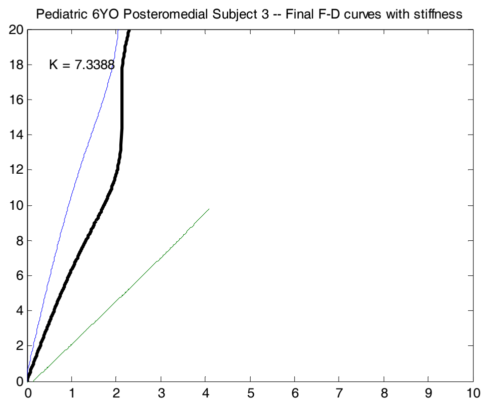


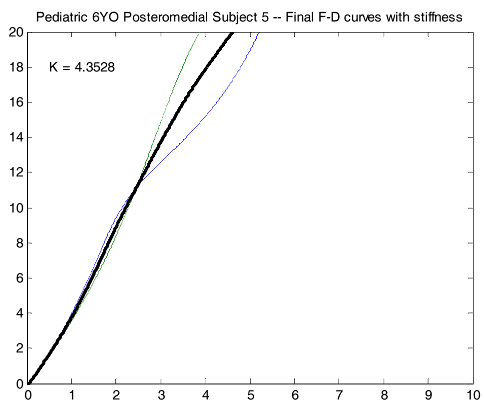


Tensed 6YO Posteromedial F-D Curves

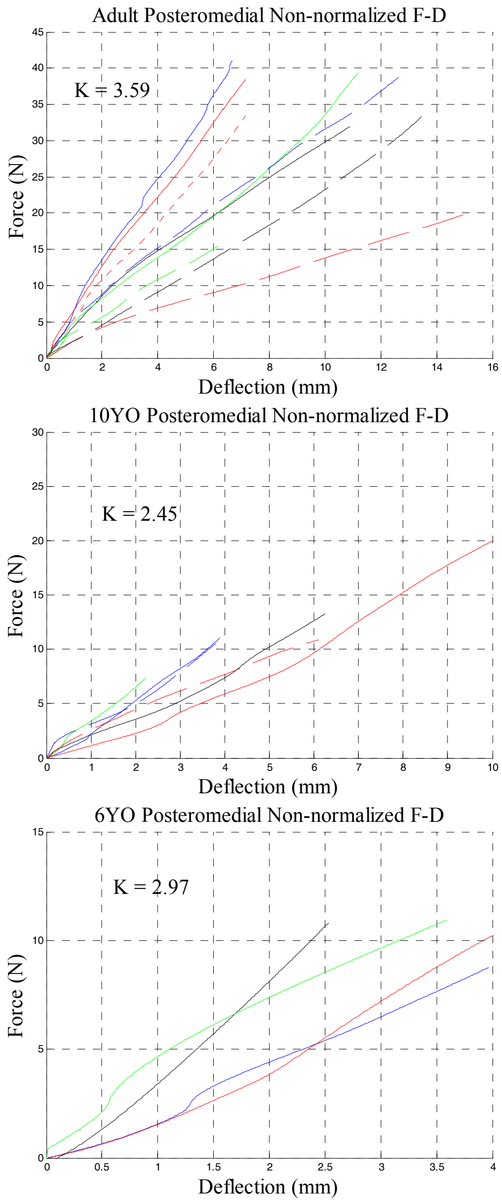








APPENDIX B



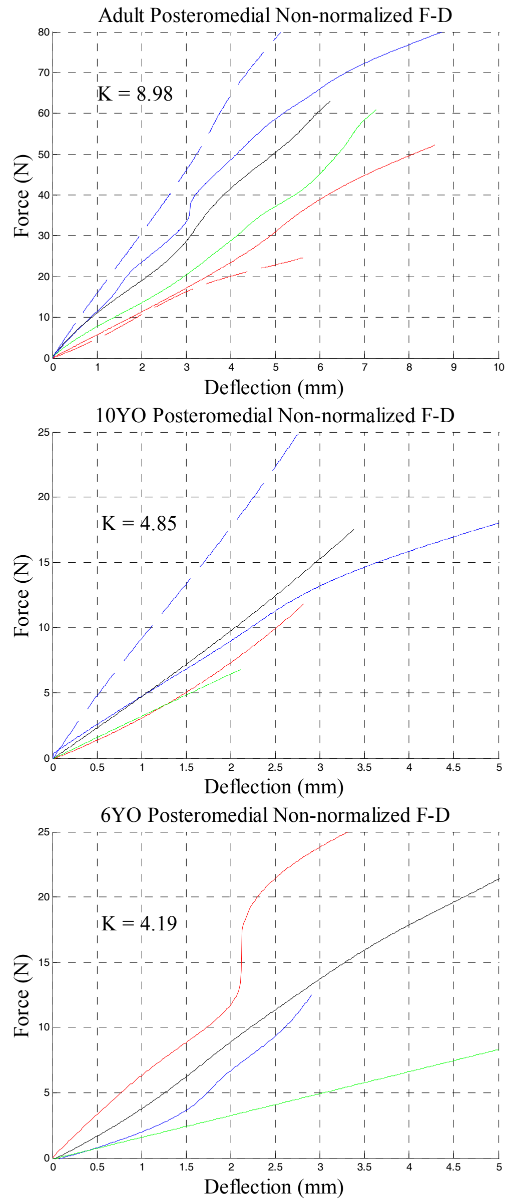


Figure B1. Non-normalized force-deflection curves for the 50th percentile adult male, 10 year old, and 6 year old in the relaxed posteromedial loading condition.

Figure B2. Non-normalized force-deflection curves for the 50th percentile adult male, 10 year old, and 6 year old in the tensed posteromedial loading condition.

APPENDIX C

Table C1.
Maximum %MVC values for a subject in each age group in both the relaxed and tensed test conditions

	Adult		10YO		6YO	
	Relaxed	Tensed	Relaxed	Tensed	Relaxed	Tensed
Latissimus Dorsi	6.29	37.29	11.62	41.15	30.92	14.05
Upper Trapezius	2.35	29.90	13.36	3.32	2.55	2.98
Anterior Deltoid	1.15	13.57	13.80	4.24	46.91	51.36
Posterior Deltoid	2.37	27.74	5.47	26.04	4.27	7.62
Biceps Brachii	1.40	10.96	3.35	3.37	4.55	3.22
Pectoralis Major	5.49	14.12	8.26	17.21	11.97	41.11
Average %MVC	**3.18**	**22.26**	**9.31**	**15.89**	**16.86**	**20.06**

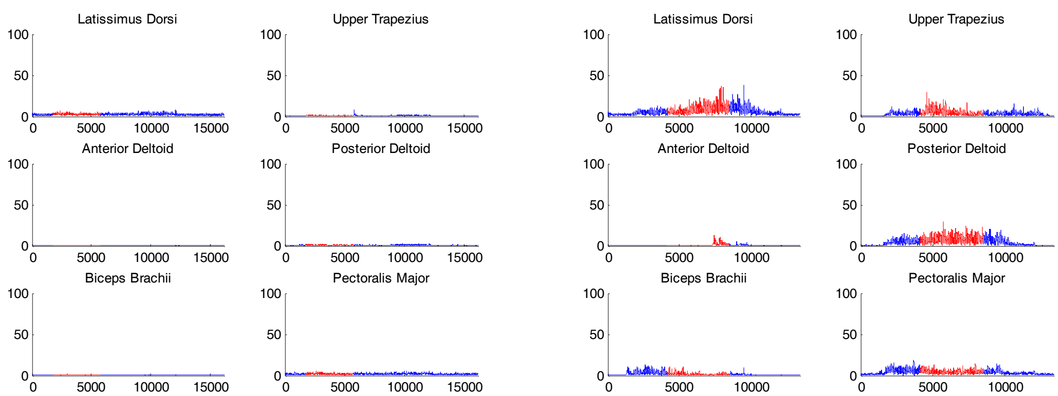


Figure C1. Plot of relaxed (left) and tensed (right) EMG signals for an adult subject. The y-axis represents a percentage of maximum voluntary contraction. The x-axis represents time (ms). The area highlighted in red is the interval in which a load was applied to the shoulder.

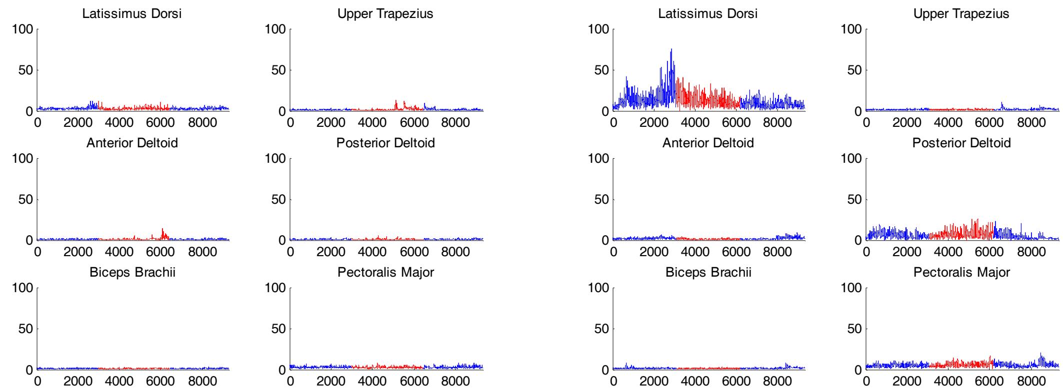


Figure C2. Plot of relaxed (left) and tensed (right) EMG signals for a subject in the 10 year old age group. The y-axis represents a percentage of maximum voluntary contraction. The x-axis represents time (ms). The area highlighted in red is the interval in which a load was applied to the shoulder.

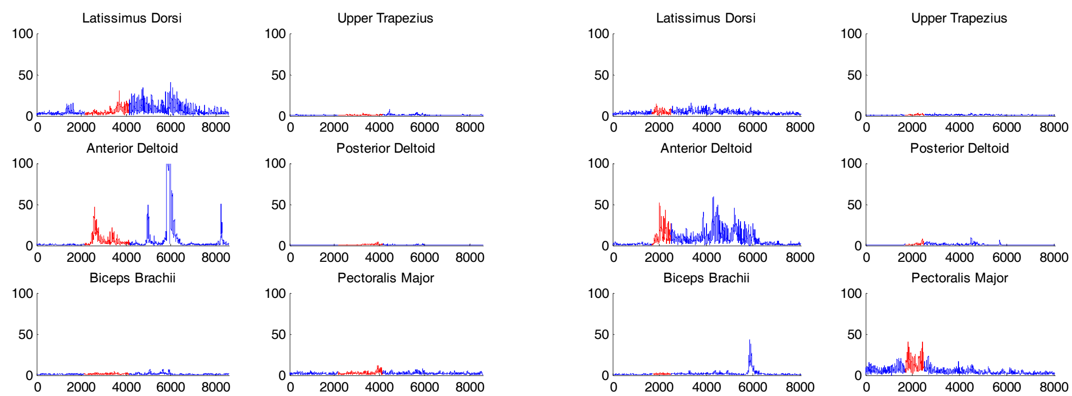


Figure C3. Plot of relaxed (left) and tensed (right) EMG signals for a subject in the 6 year old age group. The y-axis represents a percentage of maximum voluntary contraction. The x-axis represents time (ms). The area highlighted in red is the interval in which a load was applied to the shoulder.