Modeling Trajectories of Human Volunteers in Low-Speed Frontal Impacts Using Bézier Curves

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ABSTRACT

Biofidelic anthropomorphic test devices (ATDs) improve injury prevention by evaluating the effectiveness of motor vehicle safety systems. Due to a lack of pediatric validation data, pediatric ATDS have been size-scaled from adult data; but studies show that child subjects experience different motion relative to adults that are not simply scaled down by size. Previous studies note age-based differences in relative maximum excursions, but no study has quantified the effects of age on trajectory shape. In this study, low speed (<4 g) frontal sled tests were conducted on 30 human volunteers from ages 6 to 30 years. Subjects were restrained by a lap and shoulder belt. Photo-reflective markers placed on anatomical landmarks of interest, including the head top, were quantified by a 3-D near infrared tracking system. Subjects received six repetitive trials. Head top landmark trajectories in the sagittal plane were modeled using a 4th order Bézier curve equation using an iterative least-squares algorithm. A principal component analysis on the control points indicated that the first four components accounted for 95% of the variance. A linear regression analysis demonstrated that the first principal component was significantly related to subject erect sitting height (p<0.001) and other correlated measures of body size. Using the resulting regression model, shorter sitting height was associated with greater head excursion and a more rounded trajectory, indicating greater neck flexion. To our knowledge, this represents the first application of these functional shape analysis methods to impact biomechanics data. The method provides a concise, effective quantification of trajectories that will be useful for specifying and evaluating ATD performance.

INTRODUCTION

Head injuries are the most common serious injuries sustained by children in car crashes (Durbin et al. 2003). Biofidelic anthropomorphic test devices (ATDs) improve injury prevention in these situations by evaluating the effectiveness of motor vehicle safety systems. It is important that ATDs are designed such that they accurately predict the kinematics of the occupant’s head and spine motion during a crash event. Adult ATDs have traditionally been validated by comparing their response post-mortem human subjects (PMHS). However, due to a lack of pediatric validation data, pediatric ATD design has been geometrically-scaled from adult data (Irwin and Mertz 1997). However, the few studies that have compared the pediatric ATDs to pediatric (Sherwood et al. 2002) or pediatric-sized PMHS (Lopez-Valdes et al. 2009) have shown differences in head rotation, thoracic spine flexion, neck angle, head and spine accelerations, and torso angle compared to the PMHS. A recent study comparing the response of pediatric and adult volunteers in low-speed frontal sled tests suggests that geometry does not completely account for the differences between children and adults (Arbogast et al. 2009). Comparing a size-matched cohort of these volunteers from to the Hybrid III 6-year-old revealed differences in belt loads and spine excursions (Seacrist et al. 2010).

The aforementioned studies have noted age-based differences in maximum excursions, even after accounting for differences in size. However, to our knowledge, no study has quantified the effects of age or body size on trajectory shape. One method for achieving this goal is to use a functional modeling approach to quantitatively evaluate the factors that influence trajectory shape. Functional data analysis, a set of methods that address statistical problems in which the outcome measure is a function rather than a scalar, often proceed by fitting the outcomes with splines (Ramsay and Silverman 2005). Bézier curves are splines that have convenient properties for functional analysis of motion (Faraway et al. 2007, Faraway and Reed 2007). Bézier curves are defined by a set of control points that determine the shape of the curve. Regression analysis is then used to link the respective shape variables from the modeling method to covariates of interest. To this end, the current study used Bézier curves to model the trajectory of a head-top marker for pediatric and adult volunteers in low-speed frontal crashes.

METHODS

Subject Testing

A comprehensive description of the bumper car testing can be found in Arbogast, et al. (2009). Briefly, low speed (<4 g) frontal sled tests were conducted on 30 male human volunteers from ages 6 to 30 years. Subjects were restrained by a lap and shoulder belt and exposed to six repetitive trials. Photo-reflective markers were placed on anatomical landmarks of interest, such as the head top, and quantified by a 3-D near infrared tracking system. For the current analysis, the head top landmark trajectories in the sagittal plane were examined. Each trajectory was analyzed for 400 ms from the event onset. A total of 175 trajectories were included in the current analysis.

Bézier Curve Modeling

Bézier curves are defined by a set of control points, Bernstein basis polynomials and single parameter *t* which determine the shape of the curve. The first and the last control points are the start and endpoint of the curve, and the curve bends toward subsequent intermediate control points but does not pass through them (we need a figure here that shows a Bezier curve and its control points). The Bernstein polynomials are:

$$B_i^d(t) = \sum_{i=0}^{d} \left(\binom{d}{i} (1-t)^{d-1} t^i \right)$$

A Bézier curve of degree *d* is:

$$C(t) = \sum_{i=0}^{d} \left(B_i^d(t) P_i \right), t \in [0,1]$$

The Bernstein polynomials for the 4th-order curve are shown in Figure 1.

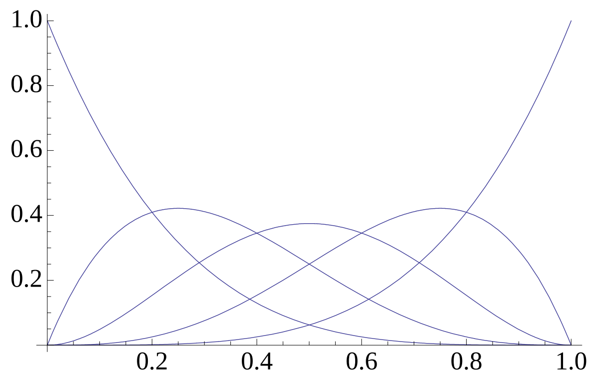

Figure 1. Bernstein polynomials as the basis functions for a 4th-order Bézier curve.

In this analysis, the trajectories were modeled using a 4th-order Bézier curve equation

$$B(t) = (1-t)^4 P_0 + (1-t)^3 t P_1 + (1-t)^2 t^2 P_2 + (1-t) t^3 P_3 + t_4 P_4,$$

where P_i are control points (X, Z) that determine the shape of the curve and the *t* [0,1] parameter represents the relative location of the current point along the trajectory. An iterative algorithm was developed that minimizes the sum of squares error (SS), where Z_i are the actual data points and $C(s_i)$ are the points on the fitted Bézier curve:

$$SS = \sum_{i=1}^{n-1} abs\left(Z_i - C(s_i) \right)^2, t \in [0,1]$$

The initial Bézier fit C(t) was calculated using a chord-length position vector s_i, used as the initial *t* vector, which accounts for varying speed during the marker trajectory:

$$s_i = \frac{\sum_{j=1}^{i} abs(Z_i - Z_{i-1})_j}{\sum_{j=1}^{n} abs(Z_i - Z_{i-1})_j}$$

The control points are then calculated from $P = [B^T B]^{-1} Y$,

with $Y_i = Z_i^T - B_0^d(s_i) Z_0^T - B_d^d(s_i) Z_n^T$.

At each iteration, the algorithm chooses a new *t* vector using the current fit to most closely match each data point.

Statistical Analysis

A principal component analysis was applied to facilitate the interpretation of the variance in the control point vector across subjects and trials. Linear regression was used to assess the relationship between the principal component scores and subject covariates. The resulting model predicts the planar landmark trajectory as a function of subject attributes such as sitting height.

RESULTS

The fourth-order Bezier curve fits captured the shapes of the marker trajectories extremely well. Figure 2 depicts the fits for the trajectories from one of the subjects over the six repeated trials. Fit quality was quantified by mean squared error (MSE), calculated as the average of the squared distances from each data point to the fitted curve. Across trials, the median MSE was 3.9 mm^2; the 5th and 95th percentile values were 0.44 and 10.9 mm^2, respectively.

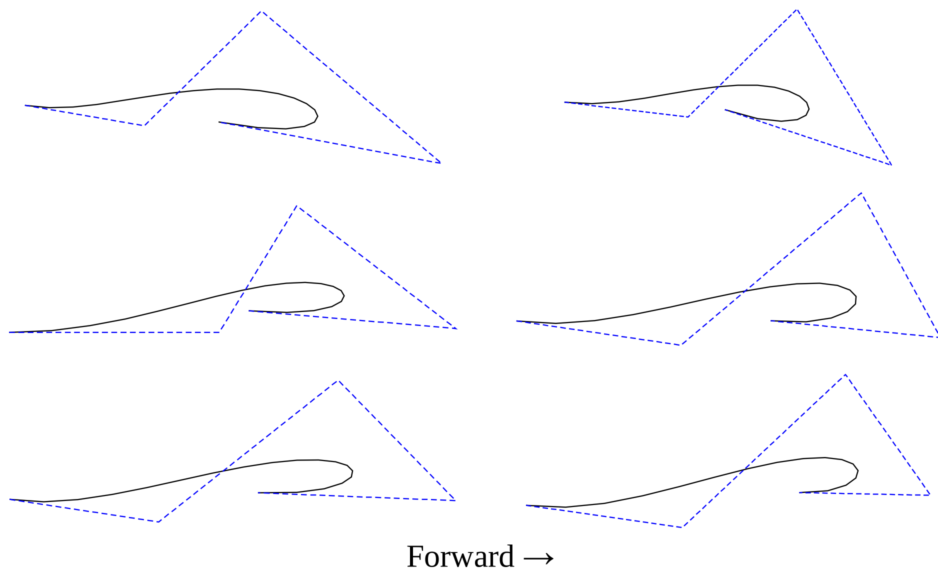

Figure 1. Bézier curve fits for the head top trajectories for one subject in six trials. Marker data are shown with dark dots; the Bézier curve fit is shown with a solid line. The dashed line connects the Bézier control points that determine the shape of the curve.

Four principal components (PC) accounted for 95% of the variance in the control point coordinates. The regression analysis demonstrated that only the first PC was significantly related to body size, using erect sitting height as the subject covariate ($p<0.001$, $R^2 = 0.89$). Figure 3 shows predictions from a regression model predicting the Bézier curves as a function of erect sitting height. On average, taller subjects (adults) had flatter and shorter head trajectories. In contrast, shorter children produced longer trajectories (greater excursion from the starting position) with larger vertical excursions, indicative of greater next flexion and head rotation.

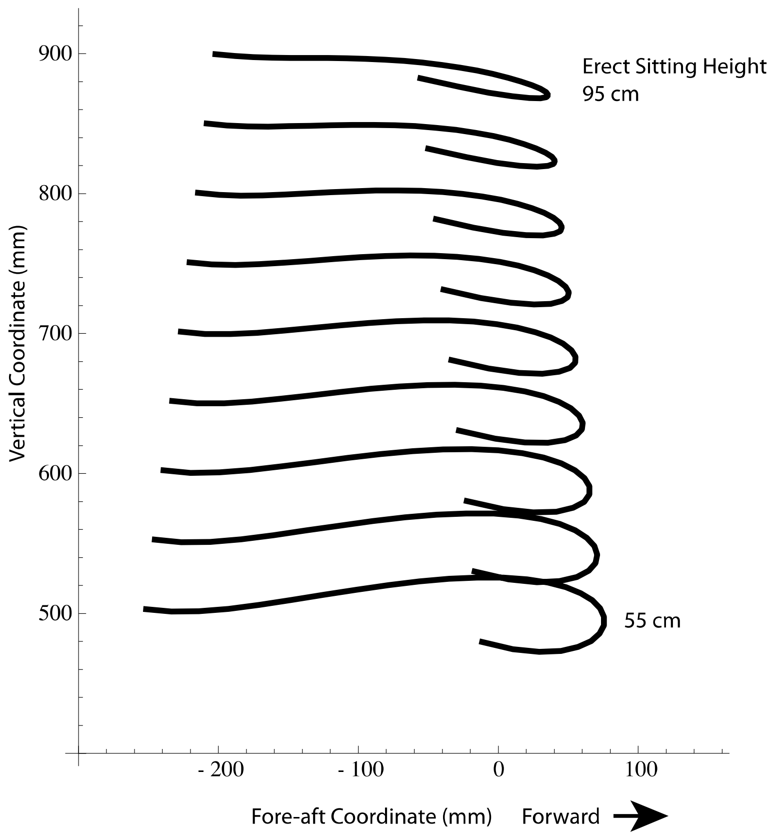

Figure 3. Effects of body size on head top marker trajectory using regression model.

DISCUSSION

This paper presents the first functional regression analysis of impact kinematics data. The results demonstrate that these planar head-trajectory data could be well represented by 4^{th}-order Bézier curves, which provide a compact and smooth representation amenable to statistical analysis. The PCA and regression analysis showed significant differences in trajectory size and shape related to body size. In particular, this statistical method allows prediction of kinematics for a particular body size (for example, children similar in size to a particular ATD) without limiting the analysis to the few children in the data set of similar size). The results are more robust to sample size limitations and provide the ability to generate confidence bounds on the predictions that can be interpreted as corridors.

The findings from the current analysis are consistent with previous research by Arbogast et al. (2009) assessing head excursion differences in pediatric and adult subjects using the same dataset. The current results emphasize the finding from the previous work that pediatric responses cannot be accurately predicted by simple geometric scaling of adult responses. Typical scaling methods predict *shorter* excursions for shorter-stature subjects, whereas these data indicate the opposite. Moreover, scaling of adult responses in the current dataset would miss the substantial differences observed in head rotation and vertical excursion between adults and children.

The current planar analysis could be extended to address 3-D motion, although additional control points would be needed. Future work will apply these functional modeling and regression analysis methods to multiple markers, and will include functional modeling methods that incorporate the temporal aspect of the trajectories.

CONCLUSIONS

This is the first study to quantitatively compare trajectory shape between children and adult volunteers in automotive-like impacts. The results in this study indicate that children experience greater forward and downward head motion, and therefore greater neck flexion, than adults, indicating that children should not be modeled as smaller adults. The method introduced in this paper provides a way to quantify the relationships between the trajectories and subject covariates, such as body size. Among other applications, the resulting statistical model can be used to specify ATD performance targets.

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