An Improved Deflection Energy Method to Normalise PMHS Thoracic Response Data

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Abstract Normalisation is the process of modifying a set of post-mortem human subject (PMHS) response data to better represent that of a standard sized human. This improved method is based on the fact that all plots of deflection energy versus deflection for the thorax are of constant slope. The deflection energy is the integral of the applied impact force over the deflection measured by a chestband up to the point of maximum deflection. Standard sized human thorax deflection energies and slopes are found from multivariate analyses relating energy and slope, separately, to the anthropometric data for all subjects. Force is the spatial derivative of the linear deflection energy curve and is a constant. The resulting rectangular force versus deflection curve leads to scale factors for force, deflection, elastic stiffness, the viscous constant, and time, assuming a two element solid viscoelastic model. Subject effective mass and post-impact velocity were calculated from conservation of energy and impulse equals momentum, solved simultaneously, providing scale factors for effective mass and subject velocity at maximum deflection. The time histories and force versus deflection have been plotted, the standard deviation targets over-plotted, and coefficients of variation calculated. Results have been qualitatively and quantitatively compared to previous methods.

Keywords biofidelity, data normalisation, post-mortem human subject (PMHS), scaling

I. INTRODUCTION

The process of normalising post-mortem human subject (PMHS) response data from impact testing to obtain a representation of the typical, or average, human response has been an important part of anthropometric crash dummy design and development for many years. Normalisation is the process of mathematically modifying the response data from a set of PMHS subjects to a standard human size.

This normalised response is used as a standard against which the response of an anthropometric dummy is compared to assess the dummy biofidelity. A quantified measure of biofidelity is desirable to provide the basis for an objective decision as to the ability of a dummy to assess vehicle crash protection for a human of similar size.

In 1984 a method for normalising PMHS data based on the ratio of the whole body mass of a subject to the *standard* total body mass (e.g. 50th percentile male) was developed [1]. This methodology assumed that all subject responses will be related directly to the whole body mass. Clearly force, deflection and kinematic responses from widely varying sizes and shapes of humans are not likely to be related solely by whole body mass. An impulse-momentum and stiffness-based normalisation method [2] was presented in 1984. Scale factors were developed from both anthropometry measures and ratios of the solution for the single degree-of-freedom (DOF) differential equation for a linear elastic system which represented a large impacting mass such as a sled type of impact. In 1989 an improvement to this methodology was presented [3] by expanding the derivation of scale factors to the two DOF linear elastic system which better represents pendulum-type impacts where the striking and struck masses are more nearly equal. Again the scale factors were developed from ratios of the solution to the system of differential equations. An improvement [4] was made to the two DOF method using the integral of the force versus deflection curve, the deflection energy, to develop an elastic effective stiffness directly from the force versus deflection response data rather than from characteristic subject dimensions. This latter study [4] also compared the effectiveness of the various normalisation methods in force versus deflection space using the standard deviation ellipse [5] and a modified coefficient of variation measure taking one half of the area of the ellipse divided by the product of force and deflection at each data point. This quantitative comparison indicated that effective stiffness collapsed the force versus deflection curves toward

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each other more effectively than the other methods. More recently the authors of this current study presented a method for normalising PMHS thoracic impact data based on the deflection energy and a two DOF viscoelastic model [6]. This approach was mathematically cumbersome but did improve the quantitative results slightly over the effective stiffness approach [4] as measured using a time based average coefficient of variation measure and the ellipse coefficient of variation measure for force versus deflection.

This study builds upon all of this prior work to develop an improved, and simpler, mechanistic method for normalising thoracic PMHS data again using a two DOF viscoelastic model. PMHS data from 19 subjects from three studies [5][7][8] of thoracic pendulum tests were analysed to illustrate the methodology. The deflection energy and the slope of the deflection energy versus deflection curve were used to develop scale factors for energy, slope, force, deflection, elastic stiffness, the viscous constant, and time. Scale factors for effective mass and subject velocity were derived from conservation of energy and conservation of momentum.

Of the previous normalisation studies the mass based method used subject anthropometry and the known anthropometry values of standard sized humans to develop scale factors for normalisation. The other earlier methods used chest anthropometry ratios and averages of the responses of the subject pool being analysed to estimate the standard sized human responses used to develop scale factors. The current study modifies this approach by using the response measures of deflection energy and slope from all 19 subjects as dependent variables and the subject anthropometry measures as independent variables to develop statistical multivariate relationships for energy and slope. The resulting equations can then be used with any standard human anthropometry values to estimate standard human deflection energy and slope values and to calculate the corresponding scale factors for the standard force, deflection, elastic stiffness, viscous constant, and time.

Results from normalising all 19 subjects to the 50th percentile male are presented, along with mean curves and standard deviation biofidelity targets, in both time history plots and force versus deflection plots. These plots have been compared with results from previous methods both qualitatively and quantitatively to demonstrate the improved grouping and smaller standard deviation targets of the deflection energy normalisation results.

II. METHODS

Response data and anthropometry data from 19 PMHS tested in lateral and oblique pendulum type impact tests in three studies were analysed using a deflection energy method and a two DOF viscoelastic model to develop scale factors for normalisation to the 50th percentile size male human. Standard values for deflection energy and the slope of the deflection energy versus deflection curve were estimated using multivariate relationships between anthropometry measures and impact velocity as independent variables and deflection energy and slope of deflection energy as dependent variables. Mechanical values for energy, force, deflection, stiffness, the viscous constant, and time were calculated from the data using the two DOF model. Effective mass and subject velocity values were found using the equations for conservation of energy and momentum solved simultaneously. Normalisation scale factors take the form

$$\lambda_p = \frac{parameter_{50th}}{parameter_{subject}} \tag{1}$$

and are applied to the relevant parameter at each point in the time history.

A description of the subjects and the data analysed, the derivation of the subject mechanical values, the development of the multivariate relationship for the standard human mechanical values, and the development of scale factors is presented below.

Subject Data

Nineteen PMHS from three thoracic impact studies were analysed in this study. Reference [6] tested seven PMHS in nominal 2.5 m/s lateral and oblique thoracic impacts at the fourth intercostal space using a 23.86 kg impactor with a 15.24 cm diameter circular face. Reference [7] tested 12 PMHS in 4.5 m/s and 5.5 m/s lateral and oblique thoracic impact tests at the xyphoid using a 23.99 kg impactor with a 15.24 cm high x 30.48 cm wide rectangular face. Reference [8] also tested five additional PMHS in 2.5 and 4.5 m/s in lateral and oblique thoracic impact tests at either the fourth intercostal space or the xyphoid using either the circular face or rectangular face impactors. Of these 24 PMHS, 19 were selected for analysis in this study. The five that were not used were eliminated for various reasons such as the impulse and momentum calculated at the time of maximum thoracic deflection being more than 30% different [5][9]. Many of these subjects were tested multiple times but only the first test was used in this study. The 19 subjects included were judged to have very good quality data for analysis and would lend confidence to a demonstration of the procedure.

The applied force versus time and the chestband deflection versus time were crossplotted and integrated up to the point of maximum deflection to obtain the deflection energy. For reasons of consistency time zero was set at the point of first continuous positive chestband deflection for every subject. Generally there was some applied force before the chestband registered deflection due to contact with the thorax skin and subcutaneous fat. As a result, at time zero there usually was an initial positive force. The alternative would be to set time zero at the initial positive force but in that case there would be no measured deflection at that time.

The data used in this analysis included lateral and oblique tests as well as circular and rectangular impact faces. Reference [8] tested a measure of elastic stiffness and indicated no significant difference in lateral versus oblique response among the various tests except for tests with the circular impact face. The standard t-test probability of α =0.05 was used in that analysis to protect against type I error and appropriately rejected the null hypothesis that the difference in the means of the stiffness was zero. In this study it was desirable to have a large set of subjects to demonstrate the method. The standard for assuming the subjects are from the same population was less stringent than in a study developing biomechanical response targets. The data was tested comparing the deflection energy curve slopes of the lateral tests to the oblique tests and circular impactor face tests to the rectangular impactor face tests in all combinations (see Appendix A). The null hypothesis in each case was that the difference in the slope values was zero. The p-values for slope varied from 0.4 to 0.9 indicating it was not possible to reject the null hypothesis of the slopes being the same. It was felt that for the purpose of demonstrating the method it was acceptable to consider all of the tests as being from the same population. This is not a contradiction of the previous result but rather presents a different view of the data, examining a different variable, for the purpose of demonstrating the normalization method. If the biofidelity response of a dummy in a particular test configuration was being assessed one might choose to use a subset of the data.

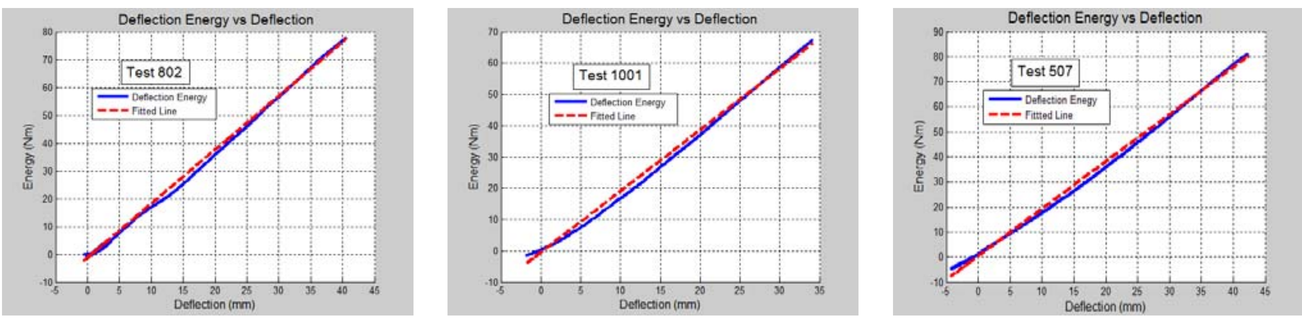
Two Degree of Freedom Viscoelastic Model

The maximum deflection energy is the integral of applied force with respect to thoracic deflection.

$$E_{max} = \int_0^{S_{max}} F(t)dS(t) \tag{2}$$

where
E=deflection energy
F=applied force
t=time
S=thoracic deflection from chestband.

The deflection energy versus deflection curves for all of the thoracic impacts examined in this study are virtually straight lines of constant slope up to, and including, the maximum deflection. Fig. 1 shows three examples of this phenomenon which are typical of all 19 subjects.



(a) (b) (c)

Fig. 1. Plots of integrated deflection energy versus deflection demonstrating the straight line slope.

In this study the term deflection energy is used indicate the energy absorbed and dissipated during the thoracic impact event up to the point of maximum deflection. Deflection energy is a one-dimensional analog to strain energy [10] where the spatial derivative of deflection energy is force rather than stress. If the deflection energy versus defelction curve has a constant slope and applied force is the spatial derivative of energy, then applied force is constant. The slope of the curve is

$$slope = b = \frac{E_{max}}{S_{max}} = F_{constant}\,.$$

and therefore

$$S_{max} = {}^{E_{max}}\!/_{slope}. \tag{3}$$

We have a relationship among energy, constant force, and deflection that can be used to develop scale factors for normalisation.

This relationship also leads directly to a two DOF viscoelastic model that can be used to find the elastic stiffness, viscous constant, damped frequency and the period. The force versus deflection curve would have constant force up to the maximum deflection as seen in Fig. 2.

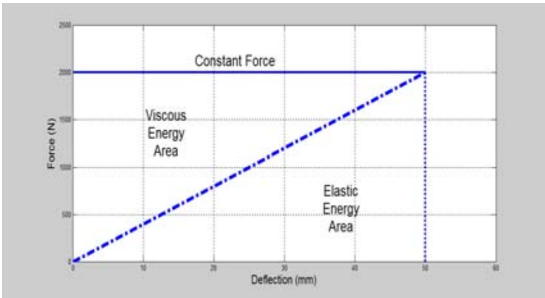


Fig. 2. Schematic of a constant force vs. deflection curve with elastic and viscous energy areas labeled.

The total deflection energy is the area under the curve, $E=F_{constant}*S_{max}$. The elastic energy portion (stored) of the energy is the lower right triangle, $\frac{1}{2}*E= \frac{1}{2}*F_{constant}*S_{max}$. The viscous energy portion (dissipated) is the upper left triangle and has the same area. The equation for the force in a two DOF viscoelastic model is

$$F(S) = K * S + C * \dot{S} \tag{4}$$

where

K = elastic stiffness

C = viscous constant

$\dot{S}$ = time derivative of deflection (velocity).

The elastic energy is

$$\tfrac{1}{2}E = E_{elastic} = \tfrac{1}{2}K * S_{max}^2 \tag{5}$$

and

$$K = \frac{E}{S_{max}^2} \tag{6}$$

The viscous energy is

$$\tfrac{1}{2}E = E_{viscous} = \tfrac{1}{2}C\dot{S}_0 S_{max} \tag{7}$$

where $\dot{S}_0$ = velocity at zero deflection or V_0

and

$$C = \frac{E}{\dot{S}_0 S_{max}} \tag{8}$$

The damped natural frequency from the differential equation for a two DOF viscoelastic solid is

$$\omega = \sqrt{MK - \tfrac{1}{4}M^2C^2} \tag{9}$$

where

ω = damped natural frequency

M = (m_I+m_s)/(m_I*m_s) (10)

m_I = impactor mass

m_s = subject effective mass

and the period is $T = 2\pi/\omega$. (11)

It should also be noted that because the viscous energy is linearly dependent on the rate of deflection, the rate must be a straight line in S equal to the impact velocity, V_0, when deflection is zero and equal to zero at maximum deflection,

$$\dot{S} = V_0 - \frac{V_0}{S_{max}}S \tag{12}$$

where

$\dot{S}$ = rate of deflection

V_0 = impact velocity

S_{max} = maximum deflection

S = deflection.

For a constant force the differential equation above is equivalent to the differential equation for the two element viscoelastic solid (Equation 4). Solving the differential equation for deflection, *S(t)*, results in an increasing exponential function in time [11]. The rate, $\dot{S}(t)$, is a complementary decreasing exponential function. When multiplied by their respective coefficients and combined the result is a constant force as is required by the constant slope of the deflection energy versus deflection curve.

The effective mass of the subject and the subject velocity at the time of maximum deflection can be found by solving the equations for conservation of energy, including the deflection energy, and conservation of momentum simultaneously from time zero to the point of maximum deflection. The resulting equations are

$$M_{effective} = \frac{(M_{impactor}^2 * V_{impact}^2)}{(M_{impactor} * V_{impact}^2 - 2 * E_{deflection})} - M_{impactor} \tag{11}$$

and

$$V'_{subject} = \frac{\left(M_{impactor} * V_{impact}^2\right) - 2 * E_{deflection}}{(M_{impactor} * V_{impact})} \tag{12}$$

where

$M_{effective}$ = effective mass of the subject

$M_{impactor}$ = impactor mass

V_{impact} = impact velocity, V_0= $\dot{S}_0$

$E_{deflection}$ = deflection energy

$V'_{subject}$ = velocity of subject at maximum deflection.

Finally, it should be noted that this approach is applicable to a two degree of freedom system (pendulum impact) or to one degree of freedom systems such as sled or drop tests by modifying the size of the impacting mass, i.e., an infinite mass for a drop test and a very large mass for a sled test.

Multivariate Regression to Estimate 50th Percentile Deflection Energy and Slope

We now have a model that provides values for energy, slope, force, deflection, stiffness, the viscous constant, time, effective mass, and subject velocity for each subject that can be used to develop scaling ratios to normalise the response to that of a standard human. The standard human response was estimated from a multivariate relationship developed using the deflection energy and the slope of the energy curve, respectively, as dependent variables and the anthropometry measures for all subjects as independent measures. The Matlab function *stepwiselm* [12] was used to develop this relationship. The analysis using the *stepwiselm* function considered all of the independent anthropometry variables available from the three studies and eliminated those variables that were not significant leaving an equation for the dependent variable based only on the most important variables and the interactions. This procedure is generally referred to as *stepwise forward estimation* [13]. The considered variables were gender, age, height, mass, body mass index BMI, chest breadth, chest depth, and impact velocity. The two multivariate equations found for energy and slope were based on subject age, mass, height, BMI, and impact velocity. The anthropometric data for the standard human (50th percentile male), along with the desired impact velocity, were then input to the multivariate equation to obtain energy and slope values for the standard human. The standard human energy and slope values and the equations derived above provide the standard human values for force, deflection, stiffness, the viscous constant, and time. The standard human energy is used in the equations for conservation of energy and conservation of momentum to solve for the standard effective mass and subject velocity at maximum deflection.

Normalisation Factors

Ratios, shown in Table I, made up of the standard parameter values relative to the subject parameter values provide normalisation scaling factors for each subject that were used to normalise the response time histories and force versus deflection curves to that of the standard human. Impact velocity is a linear coefficient in the differential equations for the two DOF viscoelastic model and, therefore, velocity was scaled before calculating the normalisation factors in Table I, i.e., the applied force, chestband deflection, and the equations for conservation of energy and momentum were all scaled for impact velocity of a common value. The 19 subjects analysed here were tested at nominal impact velocities of 2.5, 4.5 and 5.5 m/s (as shown in Table II in the Results section). To demonstrate the method the normalisation impact velocity was selected to be 3.5 m/s for all 19 subjects in this analysis. The multivariate equation was developed using the actual impact velocities but when estimating energy and slope values for the 50th percentile male the normalisation value of 3.5 m/s was input.

Table I.
Normalisation scale factors.

Impact velocity	Energy	Slope	Force	Deflection	Stiffness	Viscous constant	Time	Effective mass	Subject velocity
$\lambda_0 = \frac{V_{test}}{V_0}$	$\lambda_E = \frac{E_{50}}{E_s}$	$\lambda_b = \frac{b_{50}}{b_s}$	$\lambda_F = \frac{F_{50}}{F_s}$	$\lambda_S = \frac{S_{50}}{S_s}$	$\lambda_K = \frac{K_{50}}{K_s}$	$\lambda_C = \frac{C_{50}}{C_s}$	$\lambda_T = \frac{\omega_s}{\omega_{50}}$	$\lambda_m = \frac{m_{50}}{m_s}$	$\lambda_V = \frac{V'_{50}}{V'_s}$

Force Versus Deflection Standard Deviation Ellipses

The standard deviation ellipses can be constructed from the standard deviations for force and for deflection calculated at each point in time. In previous studies [4][14] the time based standard deviation ellipses were plotted in force versus deflection space which is misleading since the force versus deflection pairs of each subject based on the time point are not co-located in force versus deflection space. The result of this phenomenon is that the time-based standard deviation ellipses do not accurately represent the variance, and the grouping, of the force versus deflection curves in force versus deflection space. Interpolating force and deflection to a common deflection energy base resolves this problem and the ellipses can then be calculated and located accurately in force versus deflection space. However, comparing the effectiveness of this deformation energy based normalisation methodology to previous methods using the average ellipse coefficient of variation can only be done accurately if the coefficients of variation are calculated the same way. In order to make a reasonable comparison among the several normalisation methods, the coefficients of variation were calculated based on standard deviations calculated from the force and deflection time histories even though this is, strictly speaking, not correct. Since the deformation energy was not calculated for the previous methods, there was little choice. The coefficient of variation for the more accurate method was also calculated for the deflection energy normalisation method presented in this study and is both plotted and tabulated.

III. RESULTS

The test information and the anthropometric data for the 19 subjects analysed in this study are presented in Table II along with the anthropometric values and impact velocity used for the 50th percentile male human.

Table II.
Test and subject characteristics.

Subject #	Ref.	Vo (m/s)	Side Angle	Plate (cm)	Impactor mass (kg)	Gender	Age (yrs)	Height (cm)	Mass (kg)	BMI (kg/m^2)
0503	[6]	2.62	left 60^o	circle	23.86	male	79	180.3	65.8	20.2
0504	[6]	2.43	left 90^o	circle	23.86	male	80	165.1	80.7	29.6
0505	[6]	2.46	right 90^o	circle	23.86	male	77	177.8	66.2	21.0
0506	[6]	2.54	right 60^o	circle	23.86	male	87	175.3	65.8	21.4
0507	[6]	2.45	left 60^o	circle	23.86	male	53	165.1	65.3	24.0
0601	[6]	2.57	left 90^o	circle	23.86	male	63	185.4	93.0	27.0
0802	[7]	4.77	left 90^o	rectangle	23.99	male	67	169	81.6	28.6
0803	[7]	4.58	left 60^o	rectangle	23.99	male	77	178	64	20
0804	[7]	4.47	left 60^o	rectangle	23.99	male	81	159	52	20.7
0902	[7]	4.63	left 90^o	rectangle	23.99	male	60	183	89	26.5
0903	[7]	4.68	left 90^o	rectangle	23.99	female	29	175	79	25.6
0904	[7]	5.64	left 90^o	rectangle	23.99	female	81	177	79	25.5
0905	[7]	5.50	left 60^o	rectangle	23.99	male	81	175	93	30.1
1001	[7]	4.76	left 90^o	rectangle	23.99	male	82	175	72	23.5
1002	[7]	4.44	left 90^o	rectangle	23.99	male	87	186	71	20.6
1003	[7]	4.56	left 60^o	rectangle	23.99	male	48	175	78	25.5
1201	[8]	2.63	right 60^o	rectangle	23.99	male	56	176.0	88.0	28.4
1202	[8]	2.57	right 60^o	circle	23.99	male	76	183.5	97.5	29.0
1203	[8]	2.56	right 60^o	rectangle	23.99	male	71	172.0	75.8	25.6
50th	[15]	3.5					45	175	77	25.5

As mentioned previously the energy and slope values estimated by the multivariate equations are found for the nominal impact velocity of 3.5 m/s. The multivariate regression coefficients for energy and for slope are presented in Table III and were entirely determined by the anthropometric properties of the subject and impact velocity. The coefficients in Table III, columns two and three, are multiplied by the individual subject anthropometric variables listed in column one (the actual subject data is presented in Table II) to obtain an estimate of the subject deflection energy and slope. Table IV presents both the integrated and the estimated values for energy and for slope for each subject, at the actual impact velocities, as well as the difference between the two values.

Table III.
Multivariate regression coefficients for energy and slope.

Variables	Energy Coefficients	Slope Coefficients
constant	-5.0140	-352.5938
Age	0.1688	2.4432
Height	0.0288	1.8116
Mass	0.0095	-2.0358
BMI	0.1845	8.4510
impact velocity	-2.3992	43.4126
age*height	-0.0010	-0.0132
age*mass	0.0009	0.0200
age*BMI	-0.0029	-0.0662
age*impact velocity	0.0005	0.0031
height*mass	0	0
height*BMI	-0.0012	0
height*impact velocity	0.0136	-0.2389
mass*BMI	0	-0.0102
mass*impact velocity	-0.0143	0.2464
BMI*impact velocity	0.0451	-0.8103

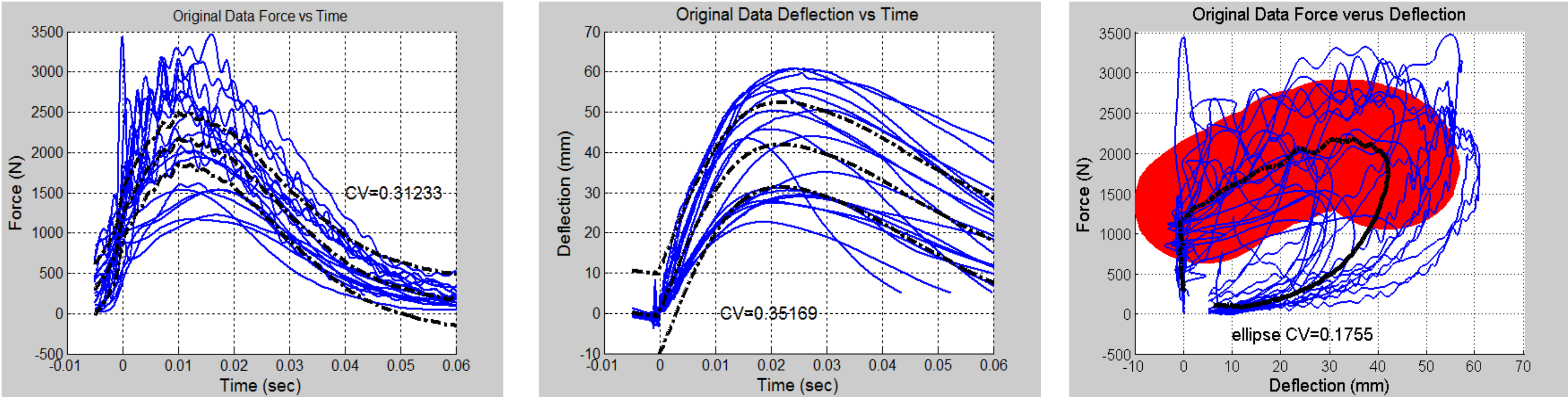
Table IV.
Integrated and estimated deflection energy and slope.

		Integrated energy & slope		Multivariate estimate		Difference			
Subject	V_0 (m/s)	E (Nm)	b (Nm/mm)	E_{est} (Nm)	b_{est} (Nm/mm)	Δ energy (Nm)	% error	Δ slope (Nm/mm)	% error
0503	2.62	32.69	1.16	32.70	1.21	-0.01	-0.03	-0.05	-4.53
0504	2.58	44.41	1.27	44.84	1.29	-0.43	-0.96	-0.02	-1.50
0505	2.46	30.00	1.04	25.66	1.04	4.34	14.46	0.00	0.15
0506	2.54	35.44	1.24	42.55	1.19	-7.10	-20.04	0.04	3.47
0507	2.45	43.27	1.00	41.56	1.01	1.70	3.93	-0.02	-1.62
0601	2.57	43.77	1.38	43.89	1.39	-0.13	-0.29	-0.01	-0.80
0802	4.77	144.71	2.64	137.57	2.58	7.14	4.94	0.05	2.01
0803	4.58	130.91	2.19	136.47	2.07	-5.56	-4.25	0.13	5.73
0804	4.47	115.44	1.94	113.47	1.93	1.97	1.71	0.01	0.47
0902	4.63	118.21	1.97	109.48	2.01	8.73	7.38	-0.04	-2.11
0903	4.68	85.31	1.74	90.15	1.66	-4.84	-5.67	0.08	4.61
0904	5.64	147.37	2.53	151.52	2.59	-4.14	-2.81	-0.06	-2.53
0905	5.50	143.94	2.58	149.06	2.55	-5.12	-3.55	0.03	1.09
1001	4.76	126.08	2.76	123.77	2.79	2.31	1.83	-0.03	-1.16
1002	4.44	108.33	2.48	104.98	2.52	3.35	3.09	-0.04	-1.51
1003	4.56	109.75	2.18	109.17	2.24	0.58	0.52	-0.06	-2.58
1201	2.63	41.60	1.75	46.85	1.83	-5.25	-12.62	-0.08	-4.49
1202	2.57	49.58	1.61	48.73	1.65	0.85	1.72	-0.04	-2.40
1203	2.56	46.77	1.65	45.16	1.62	1.60	3.43	0.04	2.13
50^{th}	2.5			55.47	1.59	**Avg. error**	-0.38	**Avg. error**	-0.29
50^{th}	3.5			81.54	1.88				
50^{th}	4.5			107.61	2.18				

Deflection energies for each subject were calculated from the test data by integrating the force versus deflection from the point in time of the initial positive increase in deflection to the maximum deflection. The slopes of each deflection energy curve were determined by dividing the maximum energy by the maximum deflection which occur simultaneously. The remaining values were calculated as described in the Methods section. The calculated values for the scale factors at the impact velocity of 3.5 m/s are presented in the Appendix.

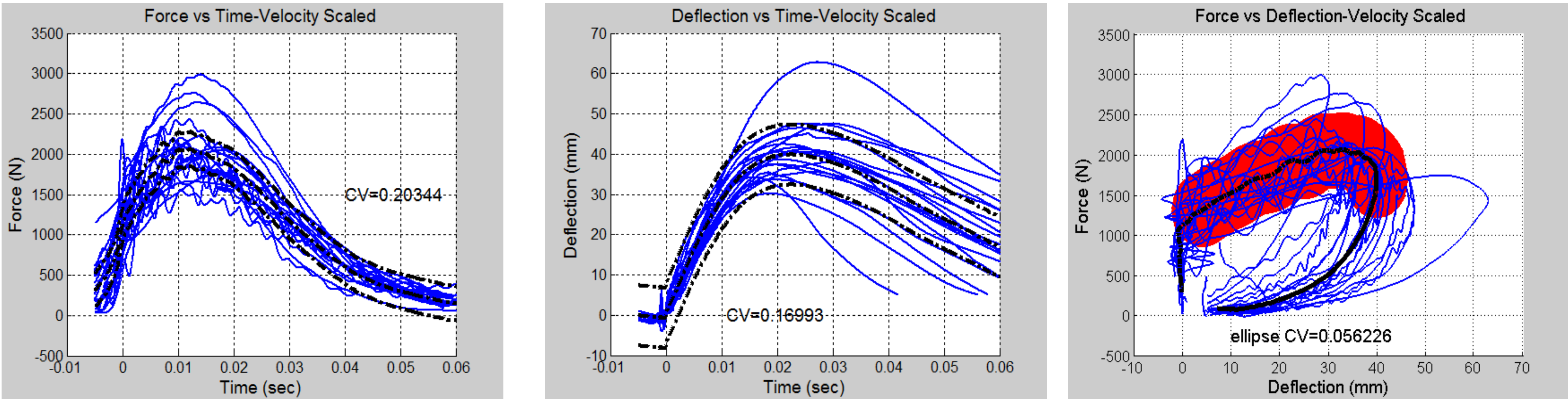
Plots of the original un-scaled and non-normalised subject force and deflection versus time and force versus deflection are presented in Fig. 3 (a-c). Plots of the same response data scaled for impact velocity are presented in Fig. 4(a-c). Plots of the response data normalised using the effective stiffness method [4] are presented in Fig. 5 (a-c). Plots of the deflection energy normalised response curves in time are presented in Fig. 6 (a-c). A plot of the deflection energy normalised response curves with the standard deviation ellipses shown in force versus deflection space is presented in Fig. 7(b) with Fig. 6(c) repeated as Fig. 7(a) for purposes of comparison.

For the time history curves the standard deviation and the CV were calculated at each time point and averaged over the entire time of the event. This approach eliminates *necking* in the plotted standard deviation time plots which is beneficial when qualitatively comparing PMHS targets to dummy responses for biofidelity assessment. The ellipse CV results were found by dividing one half of the area of the standard deviation ellipse by the product of the force and deflection at that point. The results were then averaged over the entire event to get the ellipse CV. All CVs were calculated from a starting point of 10% of the maximum deflection to the ending point of maximum deflection and direct comparisons of CVs are meaningful. The coefficient of variation results are presented in Tables V and VI.



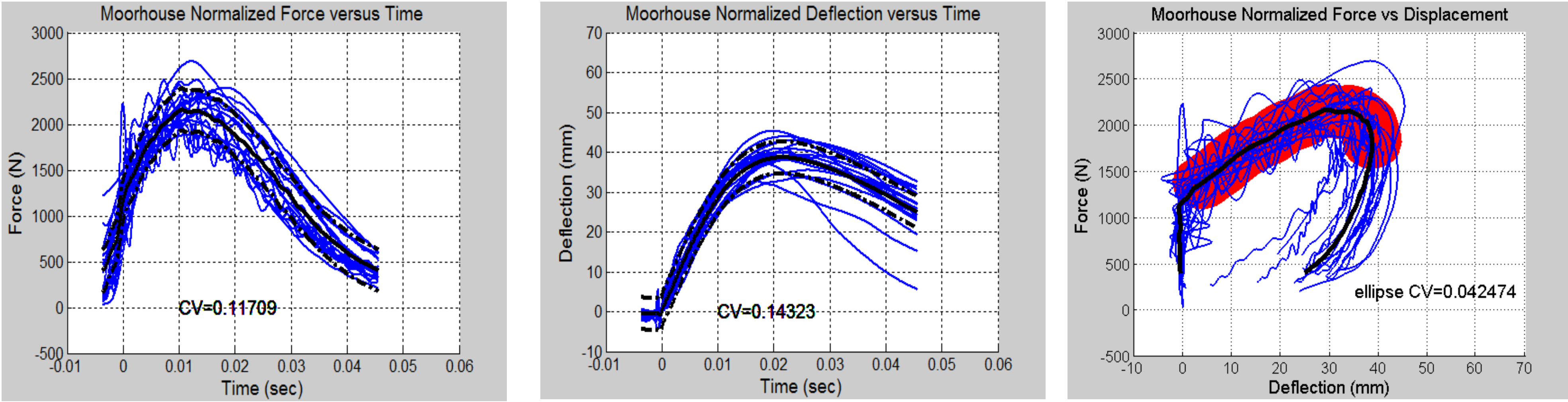
(a) (b) (c)

Fig. 3. Original response data un-scaled and non-normalised.



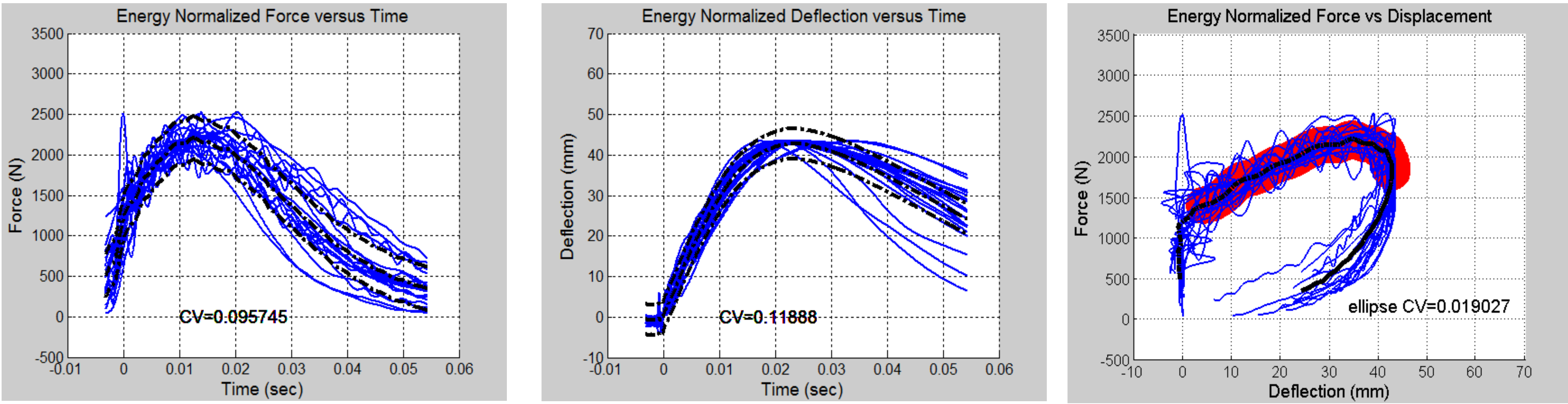
(a) (b) (c)

Fig. 4. Response data scaled for impact velocity.



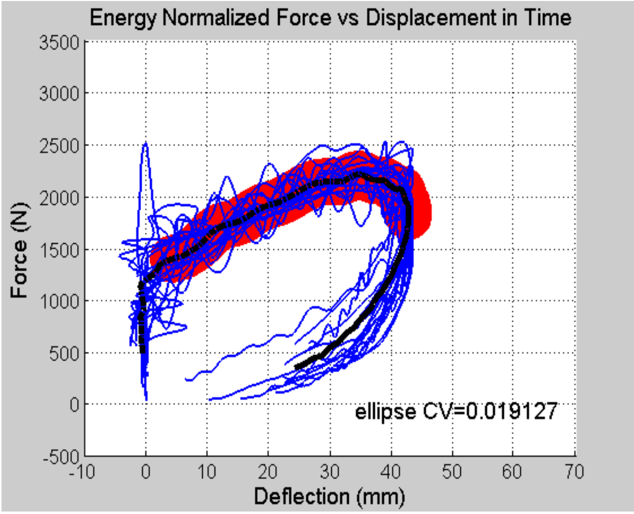
(a) (b) (c)

Fig. 5. Response data normalised using the effective stiffness method [4].

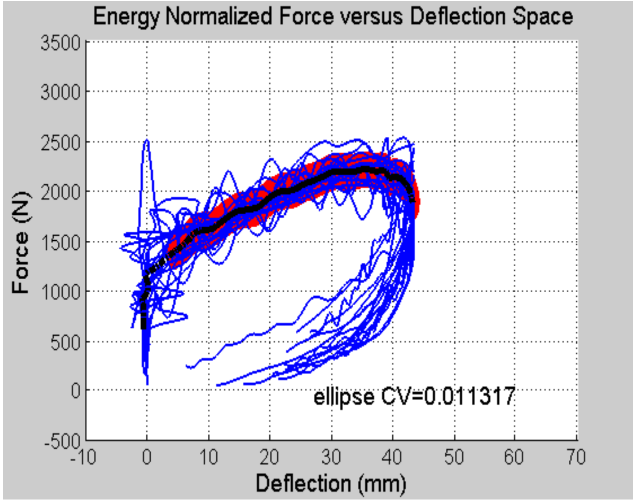


(a) (b) (c)

Fig. 6. Response data normalised using deflection energy with force versus deflection plotted using time based standard deviations.



(a)



(b)

Fig. 7. Normalised response data with (a) time based standard deviation ellipses and (b) deflection based standard deviation ellipses (note: Fig. 7(a) is Fig. 6(c) repeated).

Table V.
Comparison of average coefficients of variation calculated from time histories.

Method	Force vs Time	Deflection vs Time	Force vs Deflection
Original	0.312	0.352	0.176
Scaled for Impact Velocity	0.203	0.167	0.056
Eppinger Method	0.167	0.203	0.083
Mertz-Viano Method	0.170	0.170	0.039
Moorhouse Method	0.117	0.143	0.042
Deflection Energy Method	**0.096**	**0.119**	**0.019**

Table VI.
Average ellipse coefficient of variation calculated in force versus deflection space.

Method	Force vs Deflection
Normalised in Force versus Deflection Space	**0.011**

IV. Discussion

The objective of normalisation is to develop scale factors, based on sound mechanical principles, that are used to adjust the PMHS response to represent a standard sized human: in this example the 50th percentile adult male. It has been generally accepted that reducing the variation in a set of response data curves is successful normalisation. Previous normalisation methods assumed that PMHS subjects, and the impact responses, are normally distributed and that a set of subject response data could be used to calculate a mean response to represent a standard human (50th percentile male). Reference [2] was the first to suggest this response data approach with the effective mass approximation and Reference [4] also used it for effective stiffness. This study does not use the mean of the response data to normalise the response data but rather uses the statistical approach of multivariate regression based on the independent variables of subject anthropometry to estimate the response of the 50th percentile male. In a sense the approach used in this study harkens back to the mass based approach [1] where the response data was normalised based only on anthropometry measures. This study then developed subject response values based on a two DOF viscoelastic model and basic mechanical principles, conservation of energy and momentum, to develop normalisation factors that reduce the effect of human variation on response. The use of an elastic mechanical model to normalise response data has been used by all researchers since the mass based method and the use of a viscoelastic model is a logical step for modelling a structure with a viscous response, i.e., the human body. The assumption of normality is still used and the normalised responses are averaged and the standard deviations in time, or standard deviation ellipses in force vs deflection space, provide a target against which a dummy response can be both qualitatively and quantitatively compared when tested in the same way. Normalisation provides the appropriate biofidelity response specification for the design and assessment of standard sized dummies such that the dummy accurately models the human response.

This study improves several aspects of the deflection energy method presented in [6]. It was recognised that the slope of the energy versus deflection curve is a straight line and is equal to a constant force in a two DOF viscoelastic model. This observation makes the calculation of all necessary values for normalisation straightforward and makes the use of a viscoelastic model easily applied to normalisation of PMHS response data.

The use of the multivariate analysis approach for estimating the deflection energy and the slope of the deflection energy versus deflection curve for standard sized humans provided an independently determined set of standard human properties to be used in the scale factors. Unfortunately the data set of 19 subjects was not large enough or sufficiently well distributed enough in terms of anthropometric characteristics to effectively model the human population. In particular there were only a few smaller subjects and only two female subjects. Also a problem was the preponderance of older aged subjects. The result was a regression equation that did not model the 5th percentile female size very well. Table VII lists the estimated deflection energy for the 5th percentile female and the 50th and 95th percentile males at three ages of 45, 65, and 85 years of age. Clearly the 5th percentile female values are not useful. Fig. 8 shows the same data graphically. However, regression estimates for all of the various sized subjects in this study are reasonable as seen in Table IV. The average differences between the integrated data and the estimated values for energy and slope were -0.38% and -0.29% respectively. The maximum difference for slope was 4.61% but for energy it was -20.04%. This latter case was a smaller subject, subject 0506. The regression equation matches the modeled subject values fairly well. There is no indication of outlier subjects in the results. Nonetheless the estimate for the 5th percentile female is poor. Only the 50th percentile male estimated values were utilized in this study to demonstrate the method.

Table VII.
Estimated Deflection Energy for Several Sizes and Ages of PMHS at 3.5 m/sec.

Size/Age	45 years	65 years	85 years
5th percentile female	-99.9 Nm	58.1 Nm	216.1 Nm
50th percentile male	81.4 Nm	53.2 Nm	25.1 Nm
95th percentile male	62.0 Nm	46.5 Nm	31.0 Nm

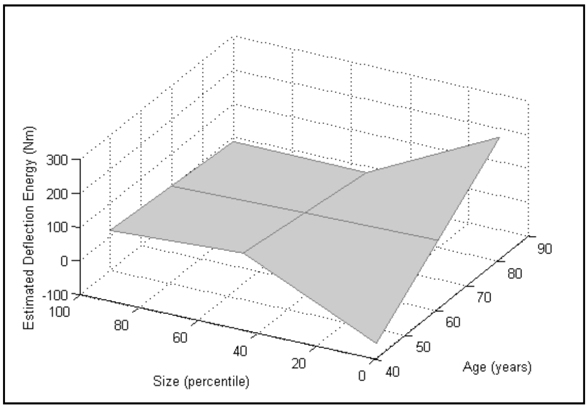


Fig. 8. Estimated Deflection Energy for Several Sizes and Ages of PMHS impacted at 3.5 m/sec.

To improve the model a larger sample of thoracic impact subjects with well distributed anthropometric characteristics would be required, particularly including smaller subjects. Those anthropometric characteristics would be the variables identified by the regression model: age, height, mass, and BMI. In addition to a linear regression model it would be interesting to examine non-linear polynomial models to better fit the data although fitting multiple variable non-linear polynomial models can be an exercise fraught with unexpected results. This study was intended to examine the use of the constant slope deflection energy versus deflection curve and the development of energy and slope estimates of standard sized humans using an approach that is independent of the response data that is to be scaled, i.e., more independent than averages of the anthropometry measures. Of course, with a large and diverse sample one would expect results from the regression model and averaging to converge.

A comparison among the previous methods was presented in [5] and plots of all results are not presented here; however, the coefficients of variation were calculated and presented in tabular form. The two-dimensional standard deviation ellipse methodology for quantifying and plotting variance was first presented in [6]. The set of standard deviation ellipses plotted at each point of the force versus deflection curve creates a target against which a dummy response curve can be both qualitatively and quantitatively assessed in force versus deflection space in the same way that time histories were evaluated using the Biofidelity Ranking System [14]. The use of the time-averaged coefficient of variation (CV) and the ellipse averaged CV for quantitative comparisons were first presented by [4] and then used by [6] to compare all of the various methods. The time-averaged CVs were again used in this study. It is clear from the plots and from the CV measures that this modified deflection energy based normalisation method eliminates a substantial amount of variation in the response data from this set of PMHS and the CVs for deflection energy normalisation are smaller than those of previous methods.

The effective stiffness method [4] used a single DOF elastic model to develop normalisation factors and found a stiffness factor by integrating deflection energy and modelling that energy as elastic energy with a constant slope. Although that study did not recognise the constant energy versus deflection slope phenomenon in thoracic impact, in effect it was utilised in that study and resulted in improved normalisation. The recognition of the constant energy versus deflection slope and the addition of a viscous response further improve normalisation as evidenced in both plots and tabulated CV values.

It can be seen in Fig. 7 and Table VI that calculating the standard deviation accurately in force versus deflection space rather than in time space reduces the ellipse sizes and reduces the CV for the data set studied here. This concept can be difficult to grasp. Consider that force and deflection are parametrically linked by time. Cross plotting the force and deflection from each PMHS test provides an accurate force versus deflection plot for each PMHS. However, the time paired set of force and deflection points for a subject are not plotted at equal deflection increments in force versus deflection space because each subject time-paired set of points occurs at different and un-equal increments of deflection. By interpolating the time paired force and deflection points onto a common deflection energy base an accurate mean curve and associated standard deviations can be calculated directly in force versus deflection space.

The essential weakness in the approach of using a relatively small sample size to characterise a large population is obvious but until a large, well-defined sample becomes available the dependence on small data sets is unavoidable. The best known model of a human is currently the PMHS and PMHS response data provides the best biofidelity target for crash dummy response. The sample size of 19 subjects tested fairly recently, all at the same laboratory (the Injury Biomechanics Research Center of Ohio State University) is a larger sample than is often available but is still too small and insufficiently diverse in anthropometry to develop a robust regression model that proves reasonable accuracy for all sizes of humans. The development of a multivariate regression model based on the independent variables of subject anthropometry and impact velocity does provide an independent and reasonably accurate estimate of the 50th percentile male human response. The approach has merit and should be pursued with a better sample of PMHS exposed to thoracic impact.

This methodology clearly reduces the variance among the subject response curves as measured by the averaged coefficients of variation for this thoracic side impact test data. Future work will examine the methodology for other body regions such as frontal thoracic impact, abdominal impact, knee-thigh-hip impact, and shoulder impact. All of these body regions have large deflections and the method may be applicable. Body regions with minimal deflection, such as the head, may not be appropriate for deflection energy normalization.

It should also be noted that an energy versus deflection curve that has a second order polynomial shape, rather than a constant slope, is still amenable to this method of developing normalisation factors. The first derivative of a second order deflection energy versus deflection curve will be a straight line, although not a constant, and the two-element viscoelastic solid model is applicable. However, the energy versus deflection curve will require two coefficients to define it and the multivariate regression process and the development of scale factors will become more complicated. It may be better to regress deflection directly in that case rather than the shape of the energy curve.

Finally, the kinematics of the impacted subject (acceleration, velocity, and displacement) can also be normalised by using the scale factors for force and effective mass to find a scale factor for acceleration and integrating to find velocity and displacement. Alternatively the scale factor for post impact subject velocity can be used by differentiating and integrating the scaled velocity to find acceleration and displacement. Note that when integrating or differentiating scaled response data the scale factor for time must also be included.

V. CONCLUSIONS

The use of deflection energy provides an improved methodology for normalising PMHS thoracic impact response data. The constant slope of the deflection energy vs deflection curve leads to the use of a two DOF viscoelastic model for development of normalisation factors. The energy and slope values for the 50^{th} percentile male human were estimated using multivariate regression. In the data analysed here this improved deflection energy method provides less variation (tighter grouping of response curves) than do the previous methods when compared using the average coefficient of variation.

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VII. APPENDIX

Analysis of Mean Responses from Oblique versus Lateral Impacts and Circular versus Rectangular Face Impacts

Results from testing the hypothesis that the means of several comparisons of the slopes of the deflection energy versus deflection curves are from the same population are shown in Table AI. The comparison pairs were a rectangular face impacting laterally and obliquely at 4.5 m/sec; a circular face impacting laterally and obliquely at 2.5 m/sec; lateral impacts with circular and rectangular faces scaled to 3.5 m/sec; and oblique impacts with circular and rectangular faces scaled to 3.5 m/sec. The null hypothesis in each case is that the difference in the slope values is zero. The p-values in each case are quite large, well above the generally accepted value of 0.05, indicating it is not possible to reject the null hypothesis. It was assumed, for the purposes of demonstrating this normalization method that the subject response data comes from the same population.

Table AI

Comparison of the slopes from oblique versus lateral impacts and circular versus rectangular face impacts.

test type	rec lat 4.5	rec obl 4.5	cir lat 2.5	cir obl 2.5	rec lat 3.5	cir lat 3.5	rec obl 3.5	cir obl 3.5
	2.64	2.19	1.27	1.16	1.94	1.72	1.67	1.55
	1.97	1.94	1.04	1.24	1.49	1.48	1.52	1.71
	1.74	2.18	1.38	1.00	1.30	1.88	1.64	1.43
	2.76				1.57		1.67	2.19
	2.48				2.03		2.33	
					1.95		2.26	
mean	2.32	2.10	1.23	1.13	1.71	1.69	1.85	1.72
std dev	0.44	0.14	0.17	0.12	0.30	0.20	0.35	0.34
degrees freedom		v=3.8		v=3.6		v=6		v=6.7
t value		t=.94		t=0.83		t=0.12		t=.59
p value		p >.4		p >.45		p >.90		p >.55

Subject Response Values and Normalisation Factors for all Subjects

The values for the 50th percentile male as determined using the multivariate predictions are shown in the first row of Table AII. The following rows contain the relevant values for each subject and the associated scale factor which is a ratio of the 50th value over the subject value, except for the time scale factor which is the inverted frequency values.

Table AII.

Scale values and normalisation factors for all subjects at 3.5 m/s impact velocity.

Subject	Energy (Nm)	Slope (Nm/mm)	Force (N)	Deflection (mm)	Stiffness (N/m)	Viscous constant (N sec/m)	Time (sec)	Effective Mass (kg)	Subject velocity (m/s)
50th	**E=81.54**	**b =1.88**	**F =1880**	**S =43.37**	**K =43346**	**C =537**	**ω =8.50**	**m =30.88**	**Vs =1.51**
0503	E=58.33 λE=1.40	b =1.56 λ_b=1.20	F = 2620 λF=1.20	S =37.33 λs= 1.16	K = 41890 λk= 1.03	C =596 λc= 0.90	ω=10.05 λt= 1.10	m =16.06 λm= 1.92	Vs =2.08 λvs= 0.73
0504	E=81.74 λE=1.00	b =1.72 λb=1.09	F = 6020 λF=1.09	S =47.52 λs= 0.91	K = 36206 λk= 1.20	C =666 λc= 0.81	ω=10.85 λt= 0.86	m =31.06 λm= 0.99	Vs =1.50 λvs= 1.00
0505	E=60.72 λE=1.34	b =1.48 λb=1.27	F = 6390 λF=1.27	S =41.09 λs= 1.06	K = 35969 λk= 1.21	C =600 λc= 0.89	ω=10.79 λt= 0.98	m =17.20 λm= 1.80	Vs =2.02 λvs= 0.75
0506	E=67.30 λE =1.21	b =1.67 λb=1.13	F = 0290 λF=1.13	S =40.41 λs= 1.07	K = 41220 λk= 1.05	C =655 λc= 0.82	ω=11.32 λt= 1.00	m =20.71 λm= 1.49	Vs =1.86 λvs= 0.81
0507	E=88.30 λE=0.92	b =1.41 λb=1.34	F = 9680 λF=1.34	S =62.77 λs= 0.69	K = 22416 λk= 1.93	C =574 λc= 0.94	ω=11.06 λt= 0.64	m =37.55 λm= 0.82	Vs =1.34 λvs= 1.12
0601	E=81.17 λE =1.00	b =1.91 λb=0.98	F = 6730 λF=0.98	S =42.44 λs= 1.02	K = 45066 λk= 0.96	C =744 λc= 0.72	ω = 9.81 λt= 0.96	m =30.56 λm= 1.01	Vs =1.52 λvs= 0.99
0802	E=77.91 λE =1.05	b =1.91 λb=0.98	F = 15630 λF=0.98	S =40.70 λs= 1.07	K = 47030 λk= 0.92	C =401 λc= 1.34	ω = 9.30 λt= 1.10	m =27.87 λm= 1.11	Vs =1.60 λvs= 0.94
0803	E=76.45 λE =1.07	b =1.65 λb=1.14	F = 7200 λF=1.14	S =46.40 λs= 0.93	K = 35516 λk= 1.22	C =359 λc= 1.49	ω = 7.27 λt= 0.96	m =26.75 λm= 1.15	Vs =1.63 λvs= 0.92
0804	E=70.77 λE=1.15	b =1.49 λb=1.26	F = 4780 λF=1.26	S =47.51 λs= 0.91	K = 31350 λk= 1.38	C =333 λc= 1.61	ω = 8.29 λt= 0.94	m =22.83 λm= 1.35	Vs =1.77 λvs= 0.85
0902	E =67.55 λE =1.21	b =1.50 λb=1.26	F = 6660 λF=1.26	S =45.15 λs= 0.96	K = 33145 λk= 1.31	C =323 λc= 1.66	ω = 8.51 λt= 0.99	m =20.86 λm= 1.48	Vs =1.85 λvs= 0.82
0903	E=47.72 λE =1.71	b =1.26 λb=1.49	F = 4070 λF=1.49	S =37.81 λs= 1.15	K = 33379 λk= 1.30	C =269 λc= 1.99	ω = 5.42 λt= 1.18	m =11.68 λm= 2.64	Vs =2.33 λvs= 0.65
0904	E=56.75 λE =1.44	b =1.60 λb=1.17	F = 9120 λF=1.17	S=35.43 λs= 1.22	K = 45215 λk= 0.96	C =284 λc= 1.89	ω = 8.13 λt= 1.28	m =15.34 λm= 2.01	Vs =2.11 λvs= 0.71
0905	E=58.29 λE =1.40	b =1.64 λb=1.15	F = 9140 λF=1.15	S =35.56 λs= 1.22	K = 46114 λk= 0.94	C =298 λc= 1.80	ω =9.35 λt= 1.27	m =16.04 λm= 1.93	Vs =2.08 λvs= 0.73
1001	E =68.17 λE =1.20	b =2.03 λb=0.93	F = 6480 λF=0.93	S =33.59 λs= 1.29	K = 60413 λk= 0.72	C =426 λc= 1.26	ω = 8.17 λt= 1.33	m =21.22 λm= 1.46	Vs =1.84 λvs= 0.82
1002	E=67.32 λE =1.21	b =1.97 λb=0.96	F = 4900 λF=0.96	S =34.21 λs= 1.27	K = 57528 λk= 0.75	C =443 λc= 1.21	ω = 7.97 λt= 1.30	m =20.72 λm= 1.49	Vs =1.86 λvs= 0.81
1003	E=64.65 λE =1.26	b =1.67 λb=1.12	F = 4960 λF=1.12	S =38.65 λs= 1.12	K = 43311 λk= 1.00	C =366 λc= 1.46	ω = 8.41 λt= 1.16	m =19.23 λm= 1.61	Vs =1.92 λvs= 0.79
1201	E=73.64 λE =1.11	b =2.44 λb=0.77	F = 4420 λF=0.77	S =30.16 λs= 1.44	K = 80980 λk= 0.54	C =928 λc= 0.58	ω=11.53 λt= 1.36	m =24.73 λm= 1.25	Vs =1.70 λvs= 0.89
1202	E =91.96 λE =0.89	b =2.22 λb=0.85	F = 2150 λF=0.85	S =41.52 λs= 1.04	K = 53352 λk= 0.81	C =861 λc= 0.62	ω = 8.32 λt= 0.98	m =41.89 λm= 0.74	Vs =1.25 λvs= 1.20
1203	E=87.42 λE =0.93	b =2.34 λb=0.80	F = 3410 λF=0.80	S =37.34 λs= 1.16	K = 62709 λk= 0.69	C =914 λc= 0.59	ω = 9.24 λt= 1.09	m =36.59 λm= 0.84	Vs =1.37 λvs= 1.11