Volunteer Bracing Strategies and Variability before Low-Speed Frontal and Frontal-Oblique Sled Tests

Devon L. Albert, Hana Chan, F. Scott Gayzik, Andrew R. Kemper

Abstract Occupants likely employ a variety of bracing strategies prior to a motor vehicle collision or impact event. However, physical and computational studies assessing the effect of bracing on occupant response typically treat muscle tone as a binary variable: relaxed or braced. It is unknown how different pre-impact bracing strategies may affect occupant responses post-impact. Therefore, the purpose of this study was to analyse the bracing strategies and bracing variability of small female and midsize male volunteers prior to low-speed frontal and frontal-oblique sled tests and to perform a preliminary analysis to determine whether the variability affected occupant kinematic responses. Each volunteer experienced four sled tests with distinct conditions and was instructed to brace with maximum effort prior to the start of the test. The forces generated at the subject-buck interfaces before the onset of sled motion showed considerable inter-subject variability and some intra-subject variability. An exploratory regression analysis indicated that differences in bracing were correlated with differences in occupant kinematics during the subsequent acceleration event. These correlations differed between the males and females. The results of this study demonstrate the need for future studies on the effect of different bracing strategies and levels on occupant response during motor vehicle collisions.

Keywords Active human body models, midsize male, occupant kinematics, pre-crash, small female.

I. INTRODUCTION

Human body models (HBMs) incorporating active musculature have been developed to predict human occupant responses and injuries under a breadth of impact scenarios [1-6]. These models are intended to compensate for the limitations of physical tests and experiments. Namely, HBMs are more efficient than physical testing when conducting parametric analyses requiring a large number of tests across a variety of impact scenarios. However, HBMs that incorporate active musculature require human validation, which should encompass a spectrum of muscle activation (from completely relaxed to completely activated). One such scenario is the braced response. Bracing or tensing can be defined as an increase in muscle activation prior to an anticipated event. Furthermore, occupants may exert additional force on any vehicle components they are in contact with, e.g. steering wheel and pedals/floor pan [7-8]. Previous studies on human volunteers demonstrated that pre-impact bracing, relative to a relaxed state, can significantly affect occupant kinematics and kinetics, particularly for lower severity events [7][9-12].

Previous efforts using volunteer testing and HBMs to evaluate the effects of pre-impact bracing on occupant response treated muscle activation as a binary response: either relaxed or braced. However, laboratory volunteer tests using sled systems and driving simulations have demonstrated variability in the magnitude of pre-impact bracing between volunteers and test conditions [8][13]. Since bracing variability was not the focus of these studies, the effect of bracing variability on any subsequent volunteer responses was not evaluated. Therefore, it is unclear whether and to what extent altering the degree of bracing will affect occupant responses and risk.

Quantification of bracing strategies, i.e. bracing more or less with certain areas of the body, and bracing variability, i.e. variation in the magnitude of bracing, in the population would provide important validation data for active HBMs as well as the opportunity to further explore the effects of bracing on occupant response. Active HBMs are well suited to perform parametric analyses using a wide range of bracing inputs in order to predict their effects on occupant kinematics, kinetics, and injury for a multitude of impact scenarios. However, the necessary

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reference data for such an analysis are largely absent from the literature. Therefore, the purpose of this study was to analyse the bracing strategies and variability of small female and midsize male volunteers prior to low-speed frontal and frontal-oblique sled tests and to perform a preliminary analysis to determine whether the variability affected occupant response.

II. METHODS

Twenty volunteers (10 female, 10 male) aged 18–28 years old (avg. 23.0 ± 2.3 years) underwent a series of low-speed sled tests. The females and males were approximately 5th percentile (156.6 ± 4.8 cm, 50.6 ± 2.4 kg) and 50th percentile (176.2 ± 2.1 cm, 76.4 ± 3.8 kg) in height and weight, respectively. Volunteer testing was approved by the Virginia Tech Institutional Review Board, USA, and each volunteer signed an informed consent form at the start of each test day. Detailed methodology for this study has been published previously [14-16].

Each volunteer underwent eight sled tests across two test days, scheduled 7–10 days apart. On each test day, the subjects experienced four tests in either a frontal (principal direction of force (PDOF) = 0°) or frontal-oblique (PDOF = 330°) orientation. Half of the female and half of the male volunteers experienced the frontal orientation on the first test day, while the other half experienced the frontal-oblique orientation on the first test day. Within a test day, the sled tests were conducted at two pulse severities (1 g and 2.5 g) and two muscle tone conditions (relaxed and braced). The sled acceleration pulses were designed to simulate an autonomous braking event (1 g) and a low severity frontal crash (2.5 g; Fig. 1). The order of the tests was the same on each day: 1 g relaxed, 1 g braced, 2.5 g relaxed, and 2.5 g braced. The analysis in this study is limited to the braced condition, producing a sample size of 80 tests across demographic groups, orientations, and pulses.

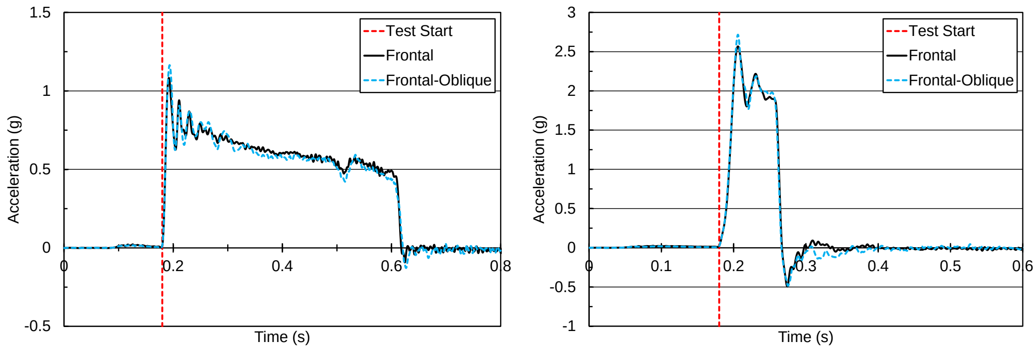

Fig. 1. Sled acceleration time histories for the 1 g (left) and 2.5 g (right) pulse severities.

A rigid test buck of previously reported dimensions was used for this study [16]. The buck consisted of a seat pan, seat back, left and right foot supports, and column with simulated steering wheel (Fig. 2.). Additionally, a standard 3 kN load-limiting, left-side, three-point seat belt from a model year 2007–2011 Toyota Camry was used for each test. The buck was originally designed for the midsize male anthropometry, so spacers were installed at each subject-buck interface to accommodate the small female anthropometry [16]. To validate these modifications to the test buck, the initial positions and weight distributions of the two demographic groups were compared in a previous study. The initial positions and weight distributions were similar between the demographic groups for both the braced and relaxed conditions [16]. The buck was instrumented with 6-axis load cells at the seat back (Denton-2513, 44 kN, Rochester Hills, MI, USA), seat pan (Denton-2513, 44 kN), and left (Denton-1716A, 13.3 kN) and right (Denton-1794A, 13.3 kN) foot pedals. The steering column was instrumented with a 5-axis load cell (Denton-1968, 22.2 kN).

The volunteers were positioned in the centre of the test buck, mediolaterally, with their feet centred on the foot pedals and hands on the simulated steering wheel for each sled test. For the braced tests, the volunteers were given a countdown from three to zero (trigger), so that they were aware of the test initiation. The volunteers were instructed to begin bracing two seconds prior to the test initiation by pushing with maximum effort using their upper and lower extremities, as if anticipating a crash event. Before conducting the tests, the researchers conducted a practice countdown and monitored the volunteers' bracing motions and timing to ensure the volunteers understood the bracing procedure.

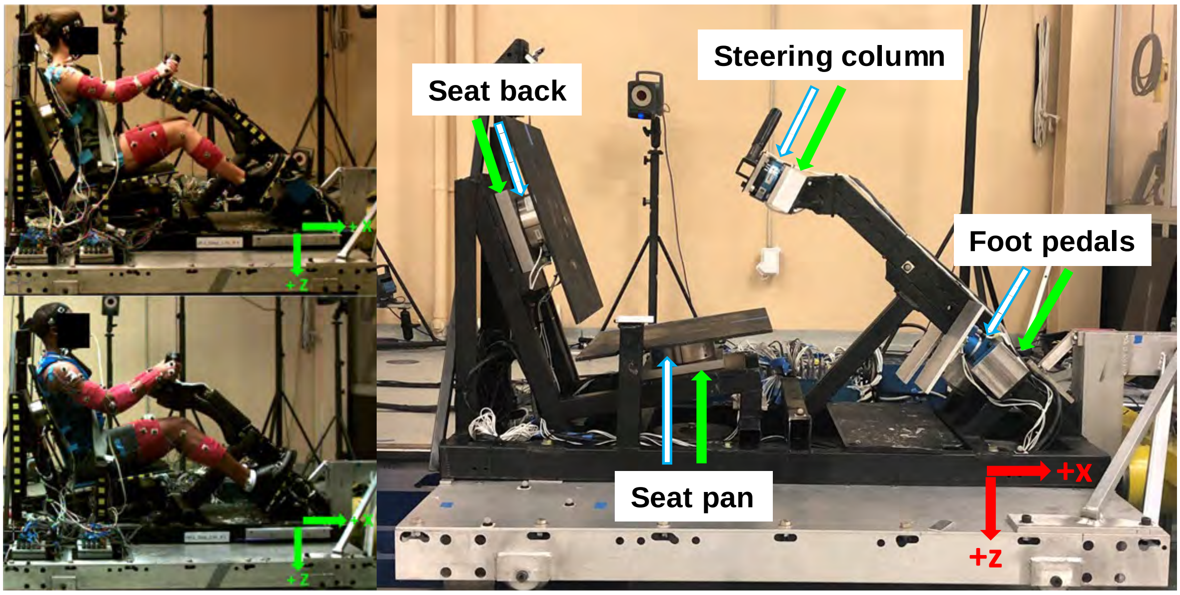

Fig. 2. Braced female (top left) and male (bottom left) volunteers and load cell instrumented test buck in the small female configuration (right). Green arrows in the left figure indicate the spacers added to accommodate the small female anthropometry. Striped blue arrows indicate the locations of the load cells.

For the purposes of this study, bracing was defined as an increase in force exerted by the subjects on the sled buck surfaces, relative to their relaxed state, prior to initiation of the sled pulse. To quantify these forces, buck load cell data were collected for 3 s prior to trigger and 3 s post-trigger to capture the volunteers at rest before pretest bracing, the pretest bracing phase, and the pulse event. Load cell data were sampled at 20 kHz and filtered at SAE-J211 channel frequency class (CFC) 60 [17]. The data were zeroed while the subjects were sitting at rest on the test buck prior to pretest bracing. The data were then time-aligned such that the beginning of the sled pulse occurred at 0.18 s. For this analysis, only the pretest loads were of interest, so the data were truncated at 0.18 s. The resultant force on each load cell was calculated over time for the pretest data. During pretest bracing, the volunteers lifted their pelvises off the seat pan, resulting in a decrease in load relative to the pre-bracing value [18]. The pre-bracing value was trivial due to zeroing the data during this time frame. Consequently, lifting the pelvis off of the seat pan produced a non-trivial negative force, leading to an increase in resultant seat pan force. Because the increased resultant force was the product of decreased load rather than a bracing force, the seat pan was excluded from further analyses of pretest bracing.

The variability in both the magnitude and timing of subject pretest bracing force, i.e., resultant force on each surface excluding the seat pan, was of interest for this study. To evaluate this, the magnitude of bracing force (henceforth: bracing) immediately prior to the test (test start bracing) and the magnitude and timing of the peak pretest bracing were quantified. The test start bracing force for each surface (seat back, column, and left and right foot pedals) was defined as the mean resultant force during the 0.2 s immediately preceding the onset of the sled pulse (-0.02 s to 0.18 s). The relative magnitudes and timings of the test start and peak bracing were then compared. Variability was quantified by computing the inter- and intra-subject standard deviations as well as minimum and maximum values of the peak bracing timing relative to test start, peak bracing force relative to test start, test start bracing forces. Intra-subject standard deviations were calculated within subjects across conditions to determine whether subjects braced consistently when given the same bracing instructions for all conditions. Pretest bracing time histories were also qualitatively evaluated for trends. Specifically, intra-test temporal trends, trends with respect to test order, and trends across buck surfaces were investigated. All parameters were tabulated separately for males and females. Some bracing forces were normalised by the average subject weight within a demographic group to facilitate comparisons between males and females.

To evaluate the potential effect of pretest bracing variability on subject kinematic response during the subsequent sled event, an exploratory statistical analysis was conducted, relating the test start bracing magnitude to the peak forward excursions of several body regions. The methodology for attaining the peak forward excursions was reported in detail in previous publications [14][15]. Briefly, subjects were instrumented with retro-reflective markers at key anatomical locations (body regions), including the head, seventh cervical vertebra (C7), shoulders, elbows, hips, and knees. A Vicon motion capture system (Vicon Motion Systems, Oxford, United

Kingdom) quantified the 3D coordinates of the markers during each test at 1000 fps. Excursions were calculated in the buck reference frame (Fig. 3) as the position of a body region with respect to its initial position at the beginning of the test.

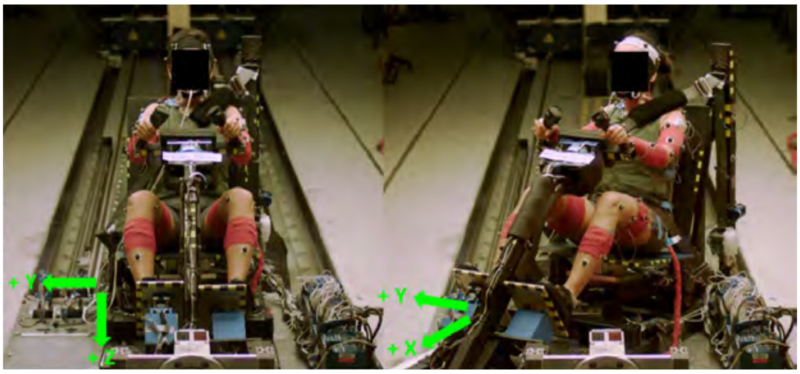

Fig. 3. The test buck coordinate system for the frontal and frontal-oblique orientations.

Differences in test start bracing between tests could not be directly compared to the subsequent kinematic responses because each sled test consisted of a unique set of conditions in terms of subject, orientation, and pulse, which could all affect the kinematic responses. Therefore, the exploratory statistical analysis consisted of a series of regression models relating test start bracing magnitude (independent variable) to the peak forward excursions of several body regions (dependent variables). Equation 1, used to represent all regression models, is as follows:

$$y = \beta_0 + \sum_{i=1}^{q}(\beta_i x_i) + \varepsilon \tag{1}$$

where y is the dependent variable, β_0 is the intercept, β_i is the coefficient for independent variable x_i, ε is the random error, and q is the number of independent variables in the model. For all models, the dependent variables consisted of the peak forward excursion of the head, C7, left and right shoulders, left and right elbows, left and right hips, and left and right knees. The independent variables for each model are summarised in Table I. The simplest model (model 1) consisted of a simple linear regression where the independent variable was the sum of the test start bracing forces across all surfaces (F Sum). Model 2 expanded model 1 into a multiple linear regression model by adding orientation and pulse as independent variables. Model 3 replaced F Sum with the test start bracing force at each surface individually, yielding a total of six independent variables.

Several metrics were used to analyse each model. The R^2 and adjusted R^2 (R^2_{adj}) were calculated to evaluate the goodness of fit of the model. An F-test was used to test the null hypothesis that none of the independent variables in the model has predictive value, as expressed in the following equation:

$$H_0: \beta_1 = \beta_2 = \cdots = \beta_q = 0 \tag{2}$$

In other words, the model as a whole has no predictive value. The p-value resulting from this test will be referred to as the model p-value. Lastly, a set of t-tests was used to test the null hypothesis that a particular independent variable has no additional predictive value given the other predictors in the model, as expressed in the following equation:

$$H_0: \beta_i = 0 \; for \; a \; given \; i \tag{3}$$

The p-values resulting from these tests will be referred to as the effect p-values or the p-value for a given independent variable, such as seat back p-value. For all statistical tests, the significance level was 0.05. Males and females were analysed separately for all models. All statistical analyses were performed using JMP Pro 16 (JMP Statistical Discovery, Cary, NC, United States).

TABLE I
INDEPENDENT VARIABLES FOR EACH REGRESSION MODEL

Model	q	Independent Variables
1	1	F Sum
2	3	Orientation
		Pulse
		F Sum
3	6	Orientation
		Pulse
		L Foot
		R Foot
		Column
		Seat Back

III. RESULTS

Four qualitative temporal trends were observed in the pretest bracing time histories: steady bracing, increasing bracing, peak before steady-state, and decreasing bracing (Fig. 4). All four trends start with an increase in bracing that begins approximately 1.5 s to 2 s before the sled pulse onset. In the case of steady bracing, the bracing force plateaus and remains relatively consistent until the beginning of the test (0.18 s). For this case, the peak bracing could occur prior to or coincident with the test start, depending on small fluctuations in force. In the event of decreasing bracing or peak before steady-state bracing, the bracing forces reach a peak prior to the test start. Then, the bracing force continuously decreases through test start (decreasing bracing), or experiences a brief period of decrease followed by a sustained constant force until test start (peak before steady-state bracing). In the case of increasing bracing, the bracing force continues to increase until test start but may experience a slower rate of increase relative to the initial ramp-up. For increasing bracing, the peak bracing will occur at the test start.

These different temporal trends resulted in differences between peak pretest bracing and test start bracing. On average, peak pretest bracing occurred 0.5 s to 0.8 s before test start, depending on the surface (Tables II–III). For almost all surfaces, the minimum timing difference observed across subjects was 0.0, indicating that peak bracing coincided with the test start. The maximum timing difference was at least 1.6 s for all surfaces. The largest difference in magnitude between test start and peak bracing was observed at the seat back. When normalised by the average subject weight, these differences represented less than 15% of subject body weight (BW) for both sexes. The inter- and intra-subject standard deviations were relatively low, representing less than 7% BW.

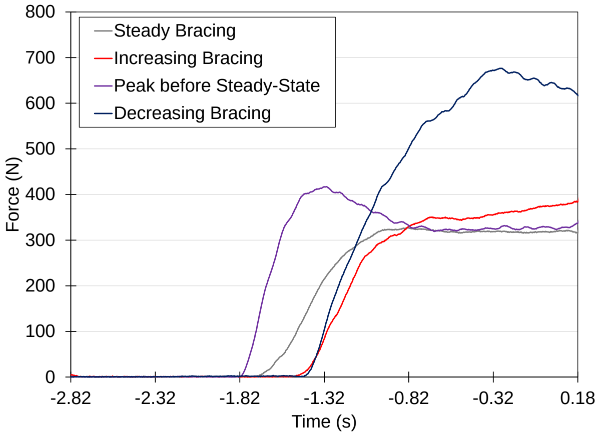

Fig. 4. Exemplar foot pedal force time histories showing four temporal pretest bracing trends: steady bracing, increasing bracing, peak before steady-state bracing, and decreasing bracing. The sled acceleration begins at 0.18s.

TABLE II

PEAK BRACING TIMING [S] RELATIVE TO TEST START FOR FEMALE (AND MALE) SUBJECTS

Measure	L Foot	R Foot	Column	Seat Back
Pooled Mean	-0.79 (-0.61)	-0.49 (-0.63)	-0.70 (-0.56)	-0.74 (-0.49)
Intra-Subject SD	0.37 (0.32)	0.40 (0.38)	0.43 (0.34)	0.32 (0.32)
Inter-Subject SD	0.19 (0.38)	0.30 (0.30)	0.33 (0.34)	0.32 (0.31)
Minimum	0.00 (0.00)	0.00 (0.00)	0.00 (-0.01)	0.00 (0.00)
Maximum	-1.60 (-1.65)	-1.65 (-1.66)	-2.42 (-1.71)	-1.82 (-1.65)

TABLE III

PEAK BRACING FORCE [N] RELATIVE TO TEST START FOR FEMALE (AND MALE) SUBJECTS

Measure	L Foot	R Foot	Column	Seat Back
Pooled Mean	24.3 (35.6)	26.3 (36.5)	35.0 (37.5)	65.8 (80.0)
Intra-Subject SD	14.0 (21.9)	14.9 (24.2)	15.0 (20.8)	33.2 (33.5)
Inter-Subject SD	14.5 (25.7)	26.8 (19.5)	20.1 (24.9)	35.3 (51.7)
Minimum	1.2 (1.8)	1.6 (1.9)	1.2 (2.4)	5.5 (9.6)
Maximum	105.2 (193.7)	121.2 (167.3)	96.9 (163.8)	217.5 (223.4)

Parameters describing the variation in test start bracing (raw and normalised by subject weight) are provided in Tables IV–V. Standard deviations of the raw force data were greater for males compared to females, but males also had larger bracing forces on average. Normalising by the average subject weight for each sex produced similar mean normalised forces and intra-subject standard deviations between sexes (Fig. 5). For both males and females, the inter-subject standard deviations were typically larger than the intra-subject standard deviations. The amount of normalised intra-subject standard deviation was similar between sexes (6–17% BW). However, the males tended to have larger normalised inter-subject standard deviations than the females (M: 15–40% BW; F: 10–30% BW). The range between the minimum and maximum normalised bracing forces for each sex was at least 50% BW for each surface.

The magnitude of pretest bracing each volunteer exerted on each surface was analysed with respect to the order in which each test was performed to evaluate whether subjects braced in a consistent manner between tests or if there were apparent trends with test order. For example, trends may include whether the magnitude of bracing for a particular test (i.e. first, second, third, fourth) was commonly higher or lower than other tests, or if the magnitude of bracing was commonly higher for one test day compared to the other. Apart from bracing consistently between tests (all tests within 50 N, n=18), the most common trends observed across volunteers and surfaces (total n = 80) were: the first test having the highest bracing magnitude (n= 15); the first test having the lowest bracing magnitude (n=13); the tests on the first test day having the highest bracing magnitude (n=8); and the tests on the first day having the lowest bracing magnitude (n=16) (Fig. 6). It should be noted that more than one trend could be observed for the same volunteer and surface. For example, first test lowest and first test day lowest were commonly observed together. Higher first test and first day bracing were more commonly observed for the male subjects, while lower first test and first day bracing were more commonly observed for the females.

TABLE IV

TEST START BRACING FORCE [N] FOR FEMALE (AND MALE) SUBJECTS

Measure	L Foot	R Foot	Column	Seat Back
Pooled Mean	191.7 (328.5)	201.2 (296.1)	234.1 (440.0)	594.8 (1040.4)
Intra-Subject SD	33.8 (64.6)	40.0 (52.0)	42.2 (81.4)	79.1 (131.1)
Inter-Subject SD	53.7 (129.3)	64.4 (113.9)	75.2 (170.4)	139.7 (284.1)
Minimum	78.1 (68.4)	105.1 (34.0)	49.5 (58.6)	360.6 (551.5)
Maximum	323.0 (634.7)	407.4 (561.1)	445.0 (859.6)	908.8 (1717.6)

TABLE V
TEST START BRACING FORCE NORMALISED BY SUBJECT WEIGHT [%] FOR FEMALE (AND MALE) SUBJECTS

Measure	L Foot	R Foot	Column	Seat Back
Pooled Mean	38.6 (43.8)	40.6 (39.5)	47.2 (58.7)	119.9 (138.9)
Intra-Subject SD	6.8 (8.6)	8.1 (6.9)	8.5 (10.9)	15.9 (17.5)
Inter-Subject SD	10.8 (17.3)	13.0 (15.2)	15.2 (22.7)	28.2 (37.9)
Minimum	15.7 (9.1)	21.2 (4.5)	10.0 (7.8)	72.7 (73.6)
Maximum	65.1 (8.7)	82.1 (74.9)	89.7 (114.7)	183.2 (229.3)

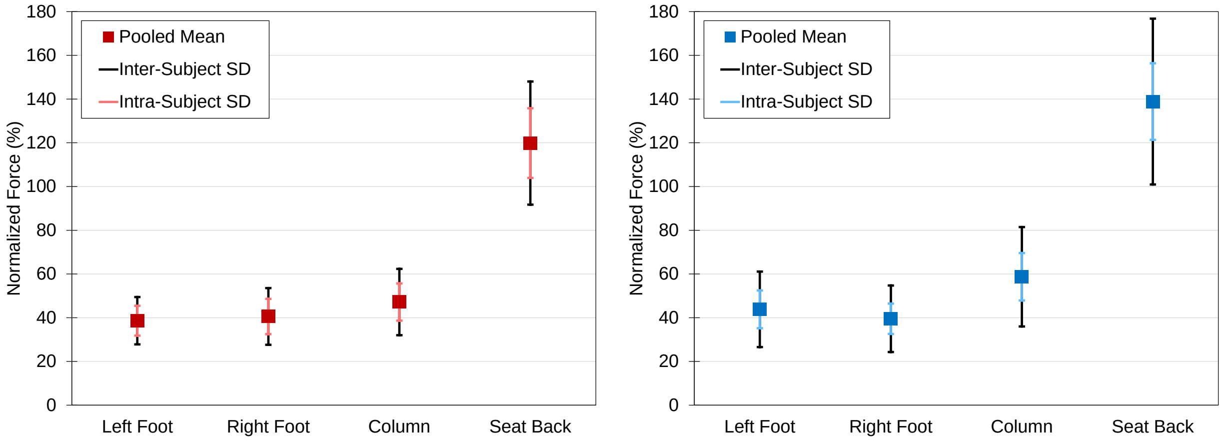

Fig. 5. Pooled means, inter-subject standard deviations, and intra-subject standard deviations of test start bracing forces normalised by subject weight for females (left) and males (right).

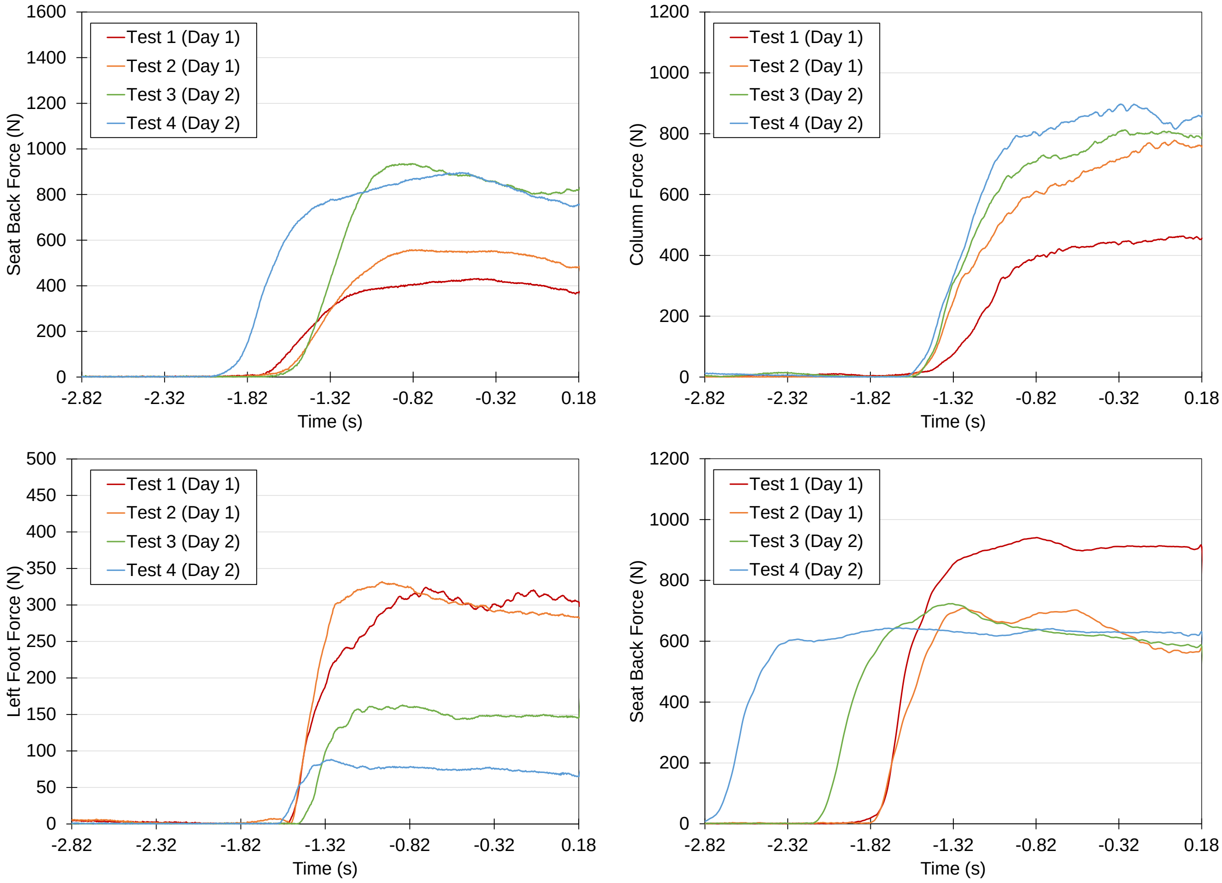

Fig. 6. Exemplar test order bracing trends: first day tests lower (top left); first test lowest (top right); first day tests higher (bottom left); first test highest (bottom right).

The exploratory regression analyses revealed that bracing force had some correlation with kinematic response. The simple linear regression model (model 1) between F Sum and peak forward excursion generally yielded low R^2 and R^2_{adj} values, indicating poor model fit (Table VI, Tables AI–AII). Despite this, model 1 produced statistically significant correlations at the left and right knees for the females, as well as the C7, left shoulder, left elbow, and right knee for the males (Tables VII–VIII). Plotting the statistically significant relationships demonstrated weak negative correlations relative to relationships that did not have statistical significance (Fig. A1–A2, Appendix). Adding orientation and pulse into the model (model 2) resulted in better fitting models with higher R^2 and R^2_{adj} values. It also increased the predictive value of the models and F Sum. For the females, F Sum had significant predictive value for the left and right hips and knees given the presence of orientation and pulse in the model. For the males, F Sum had predictive value for the C7, left shoulder, left and right elbows, left hip, and left and right knees. Breaking F Sum into its component surfaces for model 3 resulted in higher R^2_{adj} values for most relationships. However, fewer surfaces had significant predictive value given the other model effects. In fact, while all of the models had predictive value for the males, none of the bracing forces individually had significant predictive value over the other predictors in the model. For the females, some of the forces had additional predictive value, particularly for the elbow excursions.

TABLE VI
R^2_{adj} VALUES FOR EACH REGRESSION MODEL

	Female Model 1	Female Model 2	Female Model 3	Male Model 1	Male Model 2	Male Model 3
Head	-0.0249	-0.0066	0.0019	0.0292	0.0913	0.3715
C7	0.0474	0.0134	-0.0002	0.0751	0.0389	0.3315
L Shoulder	0.0036	0.3649	0.5603	0.0912	0.3560	0.5685
R Shoulder	-0.0210	0.1227	0.3241	0.0543	0.2023	0.5499
L Elbow	-0.0020	0.4019	0.6026	0.3452	0.6195	0.7524
R Elbow	-0.0238	0.4981	0.6216	0.0288	0.4408	0.6661
L Hip	0.0242	0.7123	0.7290	0.0270	0.5969	0.7097
R Hip	0.0522	0.6896	0.7184	-0.0085	0.5832	0.6177
L Knee	0.1036	0.7113	0.7586	0.0549	0.6454	0.7510
R Knee	0.1420	0.6364	0.6721	0.1272	0.6387	0.7284

TABLE VII
EFFECT P-VALUES FOR THE INDEPENDENT VARIABLES IN EACH REGRESSION MODEL FOR FEMALES
SIGNIFICANT P-VALUES ARE BOLD

	Model 1	Model 2	Model 3	Model 3	Model 3	Model 3
	F Sum	F Sum	L Foot	R Foot	Column	Seat Back
Head	0.8203	0.5921	0.1976	0.3617	0.3987	0.3206
C7	0.0946	0.1489	0.4687	0.5530	0.3150	0.5279
L Shoulder	0.2924	0.2858	**0.0379**	0.1495	**0.0096**	0.0530
R Shoulder	0.6581	0.7160	0.2905	0.4006	**0.0788**	0.2045
L Elbow	0.3428	0.1122	**0.0039**	**0.0243**	**0.0010**	**0.0073**
R Elbow	0.7602	0.3350	**0.0306**	0.1125	**0.0108**	**0.0252**
L Hip	0.1690	**0.0473**	**0.0341**	0.1584	0.0854	0.1138
R Hip	0.0840	**0.0076**	0.7465	0.2513	0.2456	0.2181
L Knee	**0.0242**	**0.0008**	0.3975	0.7955	0.8043	0.7472
R Knee	**0.0095**	**0.0005**	0.7037	0.7402	0.6274	0.5597

TABLE VIII
EFFECT P-VALUES FOR THE INDEPENDENT VARIABLES IN EACH REGRESSION MODEL FOR MALES
SIGNIFICANT P-VALUES ARE BOLD

	Model 1	Model 2	Model 3	Model 3	Model 3	Model 3
	F Sum	F Sum	L Foot	R Foot	Column	Seat Back
Head	0.1486	0.1216	0.3766	0.4468	0.5789	0.9767
C7	**0.0482**	**0.0481**	0.1459	0.2957	0.0523	0.1797
L Shoulder	**0.0327**	**0.0098**	0.8250	0.7004	0.6186	0.7646
R Shoulder	0.0798	0.0596	0.4541	0.2008	0.9057	0.3485
L Elbow	**<.0001**	**<.0001**	0.8464	0.9420	0.3626	0.8396
R Elbow	0.1501	**0.0350**	0.8273	0.8099	0.4850	0.8843
L Hip	0.1571	**0.0200**	0.6965	0.6382	0.5796	0.6996
R Hip	0.4182	0.1945	0.0599	0.2188	0.0812	0.0769
L Knee	**0.0786**	**0.0044**	0.9250	0.5097	0.7736	0.9076
R Knee	**0.0137**	**0.0002**	0.2998	0.9536	0.2989	0.2708

IV. DISCUSSION

In this study, appreciable intra- and inter-subject variation in pre-impact bracing magnitude and timing was observed despite specifying a consistent bracing target (maximum effort) for all tests and subjects. An exploratory statistical analysis demonstrated that the variability in bracing magnitude may have influenced the subsequent peak forward excursions of the subjects. Temporal trends were not evaluated in the statistical analysis, but can provide some insight into potential sources of the observed bracing magnitude variability. Two of the within-test temporal trends involved subjects reaching a peak bracing force then decreasing their bracing before test start, sometimes reaching a lower steady-state value (Fig. 4). These observations may indicate that maximum effort was not sustainable for all volunteers, contributing to more variance in the bracing magnitude at test start. Another temporal observation was that peak bracing sometimes occurred at test start, as occupants were still actively increasing the magnitude of their bracing when the sled began accelerating. This indicates that some subjects may not have had enough time to reach maximum effort. A previous study conducted in a driving simulator tracked subject bracing over time while subjects reacted to an emergency traffic event [13]. Bracing increased over time leading up to the event, but only four discrete time points were presented. Therefore, it is unknown whether or how pre-event bracing may have changed between these points. It should be noted that the difference between peak and test start bracing was relatively small, on average, for the current study (Table III). Future testing or modeling efforts would be necessary to determine whether the difference between peak and test start bracing was large enough to affect subsequent occupant responses.

Male and female subjects exhibited some differences in their bracing strategies and variance at test start. When forces were normalised by body weight, males and females typically had similar intra-subject standard deviations, but the male group had larger inter-subject standard deviations. Males and females also exhibited some differences in how their bracing magnitude changed between tests. Bracing the most for the first test was more prevalent for males (Fig. 6), while bracing the least for the first test was more prevalent among females. A possible explanation for the observed male/female differences is body/muscle mass differences. The larger males were likely capable of generating a higher voluntary maximum bracing force compared to the females, producing a larger force for the first braced test. After the braced first test, some subjects may have adjusted their bracing based on their perceived response to the event. In other words, males capable of generating higher bracing forces may have over-braced for the first test, leading to a decrease in bracing magnitude for subsequent tests. Conversely, some females may have under-braced for the first test, leading to increased bracing forces in subsequent tests. These trends should be interpreted with some caution as they were only observed consistently (across multiple surfaces) in a minority of subjects and may be representative of the strategies of individuals as opposed to whole demographic groups.

The distribution of test start bracing forces across surfaces was mostly consistent between the males and females with some small differences. When normalised to body weight, the test start bracing forces on the foot pedals were similar between males and females. Males produced higher forces at the column (10% BW) and seat back (20% BW) on average, indicating increased upper body bracing for the males relative to the females. Interestingly, the bracing forces of some female subjects showed a clear preference for upper or lower extremity bracing. Table IX shows the bracing forces across all braced tests and surfaces of interest for two female subjects. Subject X exerted considerably more force on the column compared to the individual foot pedals, while the opposite was true of Subject Y. Similar trends between upper and lower extremity bracing were not consistently observed in the males.

TABLE IX
BRACING FORCES [N] FOR FEMALE SUBJECT X (LEFT) AND FEMALE SUBJECT Y (RIGHT)

	Subject X				Subject Y			
Measure	L Foot	R Foot	Column	Seat Back	L Foot	R Foot	Column	Seat Back
Test 1	110	173	312	551	260	292	135	675
Test 2	78	130	328	496	225	258	50	518
Test 3	160	205	445	786	253	296	166	701
Test 4	142	152	380	648	264	299	125	670

As alluded to above, some of the inter-subject and inter-demographic variability may be a result of variability in muscle mass and strength between subjects. The inclusion criteria for the study controlled for height and weight, but not for factors such as body composition and strength. A previous study on male volunteer responses during low speed frontal impacts imposed strict body composition and muscle strength inclusion criteria in an attempt to minimise inter-subject variation [8]. Despite these strict criteria, the reported inter-subject standard deviations in bracing force were typically larger than those observed for the males in the current study when normalised by body weight, particularly at the seat back. It is unclear whether the volunteers in the previous study were given instructions regarding a target bracing force/effort. If not, some of the greater variation relative to the current study may have been a result of subjects self-selecting an appropriate bracing force. Even with a target of maximum bracing, appreciable intra-subject variation was observed in the current study. Subjects were not provided with quantitative feedback on their bracing and target, so it may have been difficult for subjects to reach the bracing target consistently. A previous study on volunteer head and neck responses during braced frontal and rear impacts trained volunteers to exert a bracing force equal to approximately 60% of their maximum force [11]. During the sled tests, subjects applied an average bracing force that was 57.5 ± 6.5% of their maximum, demonstrating that good consistency in bracing force resulted from this method. Therefore, providing quantitative feedback and training on bracing force appears to be a viable method of generating consistent bracing forces at prescribed levels and should be considered for future studies.

The exploratory regression analyses demonstrated that the variation in bracing levels observed in this study likely influenced subject kinematics. Model 1, the simple linear regression model, demonstrated that peak forward excursions at some anatomical locations were significantly correlated with summed bracing force for both males and females. For females, the significant correlations were limited to the lower extremity. For males, certain upper and lower extremity locations were correlated with bracing force. This model was limited because it pooled all of the test conditions together into one regression, regardless of orientation or pulse severity. This resulted in low R^2 values and poor model fit despite the significant correlations. To overcome this limitation, orientation and pulse were added to the model resulting in a multiple linear regression model, model 2. This improved the model's fit and predictive capabilities. With model 2, F Sum had significant predictive capabilities relative to orientation and pulse for all female lower extremity excursions. For the males, F Sum had significant predictive ability for peak excursions at most anatomical locations. Interestingly, all of the locations that were significantly correlated with F Sum in model 1 were also locations where F Sum was a significant individual predictor in model 2. This may indicate that F Sum was a particularly strong predictor for these locations, as it was significant in model 1 without accounting for the effects of orientation and pulse.

Model 3 incorporated the bracing forces of individual surfaces as opposed to the overall sum in order to evaluate whether certain surfaces had more impact or predictive capability for the excursions. Predictive ability of the whole model generally improved when F Sum was broken into the individual surfaces. However, excursions where F Sum was a significant predictor in model 2, did not necessarily have significant individual predictors in model 3. It was expected that statistical significance among independent variables would change as more independent variables were added. The statistical test for the independent predictors is evaluating whether each independent variable has additional predictive value once the other independent variables are considered. Therefore, this test is highly dependent on which independent variables are included in the models and their individual contributions to the model. For the females, each surface had additional predictive capability, after considering the other independent variables, for at least one of the upper extremity excursions. Given the differing lower and upper extremity bracing strategies of the female subjects discussed above, the results of model 3 may indicate that the two different bracing strategies had an effect on occupant kinematics. Conversely, the results of model 3 for the males suggested that the individual surfaces had no additional predictive ability relative to each other. Specifically, the model as a whole had significant predictive value (significant p-value, high R^2) for all excursions, but no surface or other independent variable had additional predictive value once the effects of the other variables were considered. Since the males did not seem to employ different upper and lower extremity bracing strategies between individuals, this finding appears reasonable. However, the results of model 3 should be interpreted cautiously because some of the bracing forces between different surfaces were correlated (Table AVII), violating the assumption that the independent variables were uncorrelated. Another limitation of the analyses between bracing force and subject kinematics was that each subject only experienced each test condition once. Ideally, multiple tests per condition would be necessary to better evaluate the effect of different bracing levels and strategies on kinematics. Future work will leverage active HBMs to assess how the variability in the magnitude of pre-impact bracing affects the subsequent occupant response.

V. Conclusions

This study explored the pre-impact bracing variability and strategies observed among a cohort of 20 volunteers consisting of midsize males and small females. Despite subjects receiving the same bracing instructions prior to each test, non-trivial intra-subject variation was observed. The intra-subject standard deviation ranged from approximately 7% to 17% of BW, depending on the surface. Intra-subject variation was also observed with respect to test order, where some subjects altered their bracing after experiencing the first test or test day. Inter-subject variability was generally larger than intra-subject variability, ranging from 11% to 38% of BW, and may have resulted from differences in muscle strength as well as different bracing strategies. Some male and female differences were also observed, but it cannot be determined whether these were true demographic differences or specific to the cohort tested. An exploratory statistical analysis indicated that the magnitude of the bracing variability observed in the study was correlated to differences in occupant kinematics. However, the analysis was limited because each subject experienced each test condition only once. This demonstrates the need for future studies that can directly evaluate the effect of different bracing strategies and magnitudes on occupant response. It is important to quantify pre-impact bracing variability because real-world bracing likely encompasses a wide range of bracing levels and strategies, which could lead to differences in occupant response. The data from this study can be used to inform the bracing strategies of active HBMs. Specifically, the magnitude of bracing at each surface for both small females and mid-size females could be used to simulate a fully aware, braced occupant. Additionally, the variability of bracing forces at each surface can be used in the design of future physical or computational studies assessing the effect of different bracing magnitudes on occupant response.

VI. Acknowledgements

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VII. References

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VIII. APPENDIX

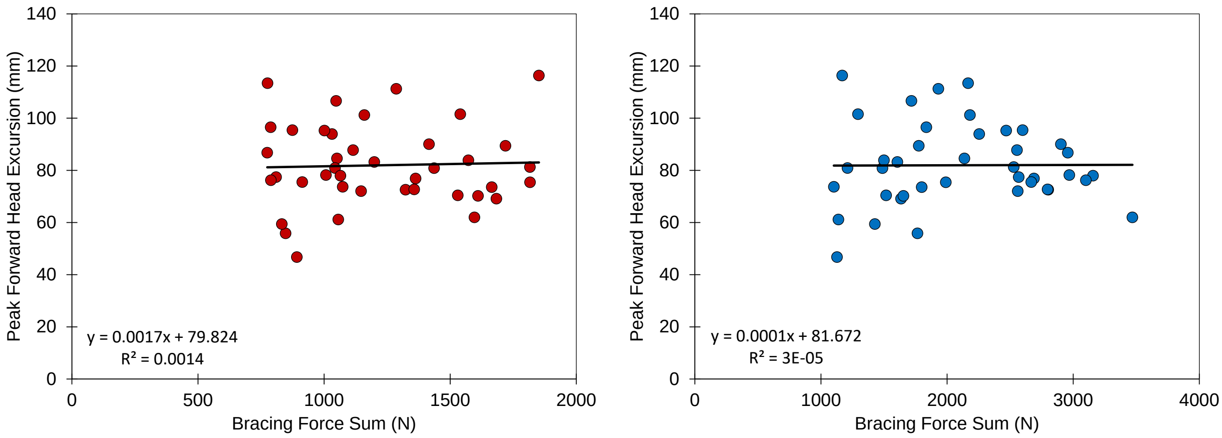

Fig. A1. Model 1 linear regression between F Sum and peak forward head excursion for females (left) and males (right).

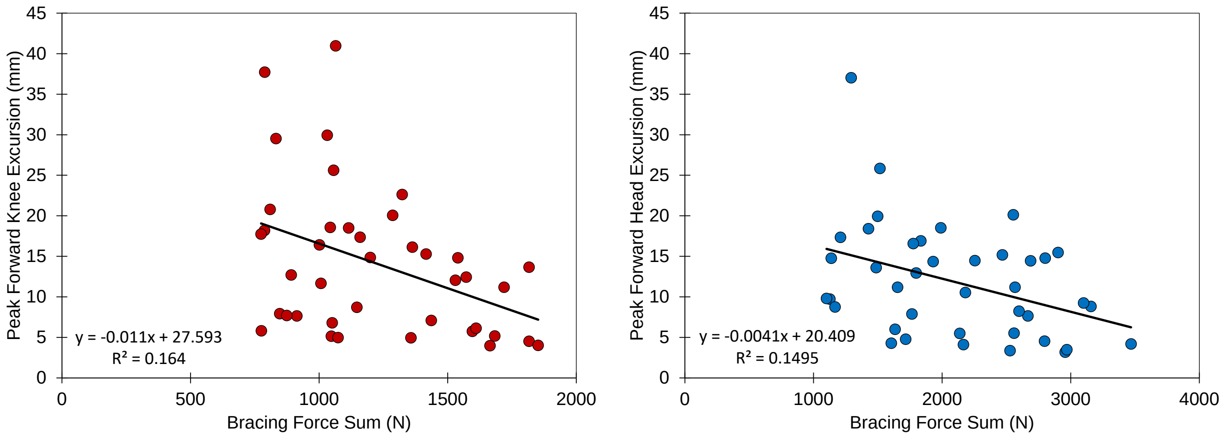

Fig. A2. Model 1 linear regression between F Sum and peak forward right knee excursion for females (left) and males (right).

TABLE AI
MODEL 1 REGRESSION PARAMETERS FOR FEMALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	p-value	Intercept	Slope
Head	0.0014	-0.0249	0.8203	79.82	0.0017
C7	0.0718	0.0474	0.0946	37.61	-0.0007
L Shoulder	0.0291	0.0036	0.2924	15.47	-0.0033
R Shoulder	0.0052	-0.0210	0.6581	13.45	0.0016
L Elbow	0.0237	-0.0020	0.3428	15.57	-0.0032
R Elbow	0.0025	-0.0238	0.7602	8.19	0.0007
L Hip	0.0492	0.0242	0.1690	28.80	-0.0073
R Hip	0.0765	0.0522	0.0840	33.28	-0.0101
L Knee	0.1266	0.1036	**0.0242**	26.38	-0.0097
R Knee	0.1640	0.1420	**0.0095**	27.59	-0.0110

TABLE AII
MODEL 1 REGRESSION PARAMETERS FOR MALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	p-value	Intercept	Slope
Head	0.0541	0.0292	0.1486	96.31	0.1486
C7	0.0988	0.0751	**0.0482**	39.76	0.0482
L Shoulder	0.1145	0.0912	**0.0327**	26.59	0.0327
R Shoulder	0.0786	0.0543	0.0798	26.06	0.0798
L Elbow	0.3620	0.3452	**<.0001**	32.67	<.0001
R Elbow	0.0537	0.0288	0.1501	12.50	0.1501
L Hip	0.0520	0.0270	0.1571	21.23	0.1571
R Hip	0.0173	-0.0085	0.4182	21.74	0.4182
L Knee	0.0792	0.0549	0.0786	19.13	0.0786
R Knee	0.1495	0.1272	**0.0137**	20.41	0.0137

TABLE AIII
MODEL 2 REGRESSION PARAMETERS FOR FEMALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	Model p-value	Orientation p-value	Pulse p-value	F Sum p-value
Head	0.0708	-0.0066	0.4437	0.1641	0.4130	0.5921
C7	0.0893	0.0134	0.3321	0.4113	0.9613	0.1489
L Shoulder	0.4137	0.3649	**0.0002**	0.6471	**<.0001**	0.2858
R Shoulder	0.1902	0.1227	0.0527	0.3963	**0.0097**	0.7160
L Elbow	0.4479	0.4019	**<.0001**	**0.0138**	**<.0001**	0.1122
R Elbow	0.5367	0.4981	**<.0001**	0.0718	**<.0001**	0.3350
L Hip	0.7344	0.7123	**<.0001**	0.3021	**<.0001**	**0.0473**
R Hip	0.7134	0.6896	**<.0001**	0.8718	**<.0001**	**0.0076**
L Knee	0.7335	0.7113	**<.0001**	0.4491	**<.0001**	**0.0008**
R Knee	0.6644	0.6364	**<.0001**	0.7471	**<.0001**	**0.0005**

TABLE AIV
MODEL 2 REGRESSION PARAMETERS FOR MALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	Model p-value	Orientation p-value	Pulse p-value	F Sum p-value
Head	0.1612	0.0913	0.0931	0.6652	**0.0428**	0.1216
C7	0.1129	0.0389	0.2242	0.6070	0.5870	**0.0481**
L Shoulder	0.4055	0.3560	**0.0003**	0.6069	**0.0002**	**0.0098**
R Shoulder	0.2637	0.2023	**0.0109**	0.5341	**0.0057**	0.0596
L Elbow	0.6488	0.6195	**<.0001**	**0.0004**	**0.0006**	**<.0001**
R Elbow	0.4839	0.4408	**<.0001**	**0.0270**	**<.0001**	**0.0350**
L Hip	0.6279	0.5969	**<.0001**	0.2669	**<.0001**	**0.0200**
R Hip	0.6152	0.5832	**<.0001**	0.5017	**<.0001**	0.1945
L Knee	0.6727	0.6454	**<.0001**	0.7891	**<.0001**	**0.0044**
R Knee	0.6665	0.6387	**<.0001**	0.5005	**<.0001**	**0.0002**

TABLE AV
MODEL 3 REGRESSION PARAMETERS FOR FEMALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	Model p-value	Orientation p-value	Pulse p-value	L Foot p-value	R Foot p-value	Column	Seat Back
Head	0.1555	0.0019	0.4340	0.1543	0.4060	0.1976	0.3617	0.3987	0.3206
C7	0.1537	-0.0002	0.4427	0.4472	0.9156	0.4687	0.5530	0.3150	0.5279
L Shoulder	0.6280	0.5603	**<.0001**	0.6563	**<.0001**	**0.0379**	0.1495	**0.0096**	0.0530
R Shoulder	0.4281	0.3241	**0.0034**	0.2962	**0.0028**	0.2905	0.4006	0.0788	0.2045
L Elbow	0.6637	0.6026	**<.0001**	**0.0021**	**<.0001**	**0.0039**	**0.0243**	**0.0010**	**0.0073**
R Elbow	0.6799	0.6216	**<.0001**	0.0553	**<.0001**	**0.0306**	0.1125	**0.0108**	**0.0252**
L Hip	0.7707	0.7290	**<.0001**	0.3181	**<.0001**	**0.0341**	0.1584	0.0854	0.1138
R Hip	0.7618	0.7184	**<.0001**	0.9492	**<.0001**	0.7465	0.2513	0.2456	0.2181
L Knee	0.7957	0.7586	**<.0001**	0.3208	**<.0001**	0.3975	0.7955	0.8043	0.7472
R Knee	0.7225	0.6721	**<.0001**	0.6315	**<.0001**	0.7037	0.7402	0.6274	0.5597

TABLE AVI
MODEL 3 REGRESSION PARAMETERS FOR MALES
SIGNIFICANT P-VALUES ARE BOLD

	R^2	R^2_{adj}	Model p-value	Orientation p-value	Pulse p-value	L Foot p-value	R Foot p-value	Column	Seat Back
Head	0.4682	0.3715	**0.0012**	0.1828	**0.0023**	0.3766	0.4468	0.5789	0.9767
C7	0.4344	0.3315	**0.0029**	0.9866	0.2386	0.1459	0.2957	0.0523	0.1797
L Shoulder	0.6349	0.5685	**<.0001**	0.6153	**<.0001**	0.8250	0.7004	0.6186	0.7646
R Shoulder	0.6191	0.5499	**<.0001**	0.3500	**<.0001**	0.4541	0.2008	0.9057	0.3485
L Elbow	0.7905	0.7524	**<.0001**	**<.0001**	**<.0001**	0.8464	0.9420	0.3626	0.8396
R Elbow	0.7175	0.6661	**<.0001**	**0.0096**	**<.0001**	0.8273	0.8099	0.4850	0.8843
L Hip	0.7543	0.7097	**<.0001**	0.0778	**<.0001**	0.6965	0.6382	0.5796	0.6996
R Hip	0.6765	0.6177	**<.0001**	0.9247	**<.0001**	0.0599	0.2188	0.0812	0.0769
L Knee	0.7893	0.7510	**<.0001**	0.9553	**<.0001**	0.9250	0.5097	0.7736	0.9076
R Knee	0.7701	0.7284	**<.0001**	0.1390	**<.0001**	0.2998	0.9536	0.2989	0.2708

TABLE AVII
LINEAR CORRELATION PARAMETERS BETWEEN FORCES INCLUDED IN MODEL 3
Significant p-values are bold

Parameters	Female p	Female R^2	Female p-value	Male p	Male R^2	Male p-value
L Foot, R Foot	0.6960	0.4844	**<.0001**	0.7567	0.5726	**<.0001**
L Foot, Column	-0.0237	0.0006	0.8847	0.1047	0.0110	0.5204
L Foot, Seat Back	0.7357	0.5413	**<.0001**	0.8115	0.6585	**<.0001**
R Foot, Column	0.0995	0.0099	0.5415	0.0003	0.0000	0.9984
R Foot, Seat Back	0.8165	0.6667	**<.0001**	0.7619	0.5805	**<.0001**
Column, Seat Back	0.5618	0.3156	**0.0002**	0.5838	0.3408	**<.0001**