Customizing Female Racing Drivers' Seat Fit Can Improve Their Performance

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Abstract

To attain peak performance, a racing driver needs to be stable in their seat. A stable driving position is thought to increase sensitivity to visual and vestibular information thereby enabling more skillful vehicle control. Optimizing seat fit to enhance stability could profoundly improve steering behavior of the driver and thus their performance. Here we report how using an electromyography (EMG)-based methodology for optimizing seat fit improved the performance of amateur women racing drivers. We measured the neck, shoulder, and trunk muscle activities of four women racing at amateur level as well as the angle of their neck and lower back with respect to vertical and their performance in terms of lap time. The seat insert decreased the neck and lower back angles of three of the four drivers. The lap times of these three drivers improved. Improved lap times were associated with changes in neck muscle activity consistent with a decrease in forward head position and reduced neck lateral flexion (head tilt). Improved lap times were also associated with changes in shoulder and trunk muscle activities consistent with adopting a right arm dominant steering action. Using electromyography to guide optimization of seat fit can have profound effects on neuromuscular processes underlying steering behavior of the driver, particularly the activity of neck muscles used to orient the head. These effects can translate to improved performance.

Keywords

Motorsports, seat fit, steering behavior, electromyography, EMG, driving position

Introduction

An optimal driving position can enhance the performance of a racing driver. Stability is critical. A stable position is thought to increase sensitivity to visual and vestibular information compared to an unstable position, thereby promoting more skillful vehicle control during cornering (Treffner et al., 2002). Training drivers to improve their stability by bracing is effective in improving vehicle control and lateral acceleration (Petersen et al., 2008). This translates to better cornering speed. However, gains in driver performance beyond those achievable through training could be generated by modifying the seat to optimize driving position.

Motorsport is unique. It is the only sport in which men and women compete against each other, ostensibly without a sex bias enshrined within the regulations (except for FIA W series). However, the reality of a sex neutral competition is far from the truth. Currently all drivers competing in the FIA Formula 1, FIA Formula 2 and FIA Formula 3 World Championships are male (*F1 Drivers 2021 - Hamilton, Verstappen, Vettel and more*, retrieved April 3, 2021; *Teams & Drivers - Formula 2*, retrieved April 3, 2021; *Teams & Drivers - Formula 3*, retrieved April 3, 2021). The male bias in the design of racing seats, restraints, and impact testing (Troxel, 2008), in racing regulations and crash testing (Welsh & Lenard, 2001), and in racing apparel (Tian et al., 2020) is likely to disadvantage female drivers in terms of performance. This male bias is particularly important in closed-cockpit racing because the design of many closed-cockpit racing cars is heavily influenced by the design of their road-based counterparts. The male bias evident in the design of road cars has long been recognized to result in women being disproportionately injured in motor vehicle accidents (Bose et al., 2011; Marshall et al., 2010; Ryan et al., 2020; Welsh & Lenard, 2001). This is thought to be due to a male bias in the design of seats and restraints (Bose et al., 2011) combined with differences in driving position (Ye et al., 2015). Consequently, women competing in motorsport are likely to be disadvantaged in terms of both performance and safety compared to their male counterparts due to key elements of vehicle design including seat design and driving position.

Recently, we reported a case study describing how we improved the performance of an experienced amateur female racing driver by using an electromyography (EMG)-based methodology to optimize her driving position (Rosalie & Malone, 2019b). The methodology that we used was first developed by Rosalie (2015) and has since been replicated by Rosalie and Malone (2018a, 2018b, 2019a). EMG optimization of seat fit resulted in a 2.18s improvement in the average lap time of the driver over a ten-lap stint. Improved lap times were associated with changes in neck muscle activity and head-neck angle of the driver that were consistent with correction of an overly forward head position. Previous work has shown that a forward head position increases lateral flexion in the direction of rotation (head tilt) resulting in increased tonic activity of the sternocleidomastoid, a pair of muscles that connect the sternum, clavicle, and mastoid process of the temporal bone and serve to turn, tilt, and nod the head (Kim, 2015). This increase in tonic activity of the sternocleidomastoid is probably due to changes in proprioceptive input from muscle spindle fibers (Pettorossi & Schieppati, 2014). Lateral flexion in the direction of rotation has the potential to cause the driver to “understeer” through corners by deviating steering trajectory away from the direction of rotation. This is because proprioceptive input from the sternocleidomastoid informs perception of rotation in the direction opposite to the anatomical location of the muscle (Bove et al., 2001; Land & Tatler, 2001; Pettorossi & Schieppati, 2014). So, activating the right sternocleidomastoid to laterally flex the head to the right during a right-hand turn will deviate perceived head angle to the left thus affecting accurate perception of steering angle. Consequently, using EMG to detect and correct coordination patterns that increase the risk of driver error caused by misperception of head angle has great potential to improve driver performance. The seat clearly influences these coordination patterns. Therefore, using EMG to guide optimization of the seat warrants further study.

This study is part of a larger investigation examining steering behavior in various categories of motorsport. In our first study we focused on drivers of open-cockpit formula cars (Rosalie & Malone, 2019a). In this, our second study, we focused on drivers of closed-cockpit cars. In upcoming studies, we will focus on motorcycle riders. The aims of the overall investigation were to investigate whether the increased attentional demands of intentionally following another car caused drivers to modify their coordination pattern for steering compared to driving on a clear track. We expected that muscle activation patterns of drivers of all three vehicle types would be affected by intentional following, but that the specific changes would depend on vehicle type (symmetrical vs. asymmetrical, motorcycle vs. car). Our hypotheses for the overall investigation were as follows: first, that intentionally following another car would result in the allocation of attention to a narrower visual search strategy resulting in a reduction of head movement and consequently a decrease in neck muscle activity; second, that a decrease in head movement would lead to a change in steering movements and thus shoulder muscle activity; third, that a change in upper limb movement would lead to a change in the activation patterns of trunk muscles.

Here we report differences in performance, muscle activity, and spinal angles for a group of female drivers who completed the task of driving on a clear track in closed cockpit cars with and without a customized seat insert. The purpose of the customized insert was to ensure that the drivers were safely harnessed to the seat, had unimpeded vision, and could comfortably operate the controls. It is standard practice to adjust the seating position of the driver in the cockpit to optimize visibility, fit of safety harnesses and position relative to the steering wheel, gear lever and peddles for safety purposes. However, this process is rarely, if ever, completed according to a validated scientific method.

Methods

Participants

The Curtin University Human Research Ethics Committee granted approval for a project using a naturalistic observational design to investigate muscle activity underlying steering behavior during practice, qualifying, and racing sessions when intentionally following another vehicle compared to driving unhindered (HR191/2014). This study focuses on closed-cockpit racing drivers. Four women gave written informed consent to participate. Driver 1 was 69 years old with 5 years' experience and the highest level of competition of club level. Driver 2 was 21 years old with 6 months experience and the highest level of competition of club level. Driver 3 was 37 years old with 18 years' experience and the highest level of competition of national level. Driver 4 was 51 years old with 33 years' experience and the highest level of competition of international level.

Location and equipment

We collected data over two weeks at a private racetrack. The 4 km (2.5 mile) long track was operated in a clockwise configuration consisting of eleven right-hand corners and eight left-hand corners.

Three drivers drove naturally aspirated Porsche 944s modified to SCCA racing specifications by Raptor Motorsports (AZ, USA). The fourth driver drove a privately owned, race prepared Honda S2000.

We measured the activation patterns of seven muscles: sternocleidomastoid, cervical erector spinae, anterior deltoid, pectoralis major, lumbar erector spinae, rectus abdominis, and transversus abdominis, bilaterally using surface electromyography (EMG) sensors with integrated inertial measurement units (IMU) (Delsys, Trigno IM, Boston, MA, USA). Data from each sensor were transmitted wirelessly to a manufacturer supplied data logger (Delsys, TPM, Boston, MA, USA) which synchronously recorded the data from the 14-sensor array. We positioned the measuring electrodes according to the recommendation of the Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles (SENIAM) Project for the placement of measuring electrodes (Hermens et al., 1999) except for the following: The electrode measuring muscle activity of pectoralis major medially was placed along the line of the sternal portion to capture its activity during downward rotation of the steering wheel (Król et al., 2007; Pick & Cole, 2006); the electrodes measuring the activities of transversus abdominis and rectus abdominis were placed according to the recommendations of Marshall and Murphy (Marshall & Murphy, 2003).

The drivers were instructed not to remove the electrodes until they had completed the testing protocol. The electromyographic data were recorded in millivolts at a rate of 1111Hz, accelerometry and gyroscopic data at 148 Hz, and magnetometry at 74 Hz. The TPM also incorporated a tri-axial accelerometer which sampled at 148Hz. We used a 10 Hz global positioning system (GPS) with an integrated 100Hz inertial measurement unit (Catapult Optimeye S5, Catapult Sports, Docklands, Australia) to measure track position and lap time. We co-mounted the Catapult S5 and Delsys TPM in the cockpit of each test vehicle and time synchronized them using a sequence of taps which were recorded by the accelerometers in each unit. The same GPS unit was used for every test which allowed an accurate determination of how muscle activity corresponded to where the drivers were on the track and how many laps they had completed.

Experimental Design and Procedures

In the days preceding our data collection, all four drivers participated in an unrelated study carried out by a separate research team on the same track using the same vehicles. The drivers were therefore likely to have acclimatized to both the track and their vehicles. We decided that it was necessary to optimize the seating position of the drivers prior to commencing the experiment proper for two reasons. The first reason was safety. An appropriate seating position reduces the risk spinal injury in the event of a crash (Trammell & Flint, 2012) . We were not confident that the generic seats installed in the cars were appropriate for the drivers. The second reason was experimental control. The drivers who participated in our previous study of steering behavior in closed-cockpit formula cars (Rosalie & Malone, 2019a) all used custom fitted racing seats. As the methodology of this study was identical to our previous study, an equivalently stable driving position was critical to reliability.

The evening before they were scheduled to drive, we adjusted the drivers' seat, pedals, steering wheel, and racing harness until they were satisfied with their driving position. In addition, a customized seat insert was molded for each driver using an SFI approved kit (BSCI Energy Impact Systems, Mooresville, NC, USA). The insert was fitted after the driver made her initial attempt at the solo task if she reported being dissatisfied with her driving position. Molding the insert involved filling a large bag, placed between the driver and her seat, with self-expanding poly-urethane foam that conforms to the position of the driver before hardening. We visually inspected the positions of Driver 1, Driver 2, and Driver 3 during the molding process to ensure that it corresponded to the position used for the case study driver (Rosalie & Malone, 2019b). Due to time constraints, the seat insert for Driver 4 was molded without our oversight.

Over the following days each driver completed her initial attempt at the solo task which consisted of completing a ten lap "qualifying session" on a clear track. Each driver commenced in the pits where she was strapped into her car. After activating and synchronizing the TPM and Catapult data loggers, each driver was asked to remain still for 30 seconds in her normal driving posture to measure baseline muscle activity and position. The driver then started her car and exited the pits.

The task was separated into three parts. First, the driver completed two warm-up laps to heat the tires and acclimatize to track conditions. After the warm-up laps, the driver immediately commenced a ten-lap stint driving as fast as safely possible. She then drove one final cooldown lap to complete the test before returning to the pits. To orient the drivers to the racing context, a green flag was waved to start the qualifying stint and a checkered flag to end it. We met each driver as she re-entered the pits to ask for immediate feedback on whether she felt secure and comfortable in the seat. None of the drivers reported being satisfied with her seating position. Therefore, a mechanic fitted the seat insert while the driver rested and rehydrated in an air-conditioned lounge. After a minimum of two hours, the driver reattempted the solo task.

Data Processing

The subset of data presented here describes the effect that optimizing driving position had on driver performance and muscle activation patterns. We processed the data using the same procedure used for the case study (Rosalie & Malone, 2019b). From the GPS unit we extracted lap time and total elapsed time. From the cervical (C7) and lumbar (L1) EMG/IMU sensors we extracted the angles of inclination of the lumbar and cervical spine with respect to gravity during the period that the driver was sitting still before commencing the solo tasks. For these data, we computed average angles over a 20s period. A negative change in angle corresponded to a more vertical position. We used the data from the rate gyroscope to confirm that the driver was still during the measurement window.

We analyzed the time-dependent median frequency of the EMG power spectrum to determine an index of muscle activity (Phinyomark et al., 2012). We imported the GPS data into Delsys EMGworks (Delsys, Boston, MA, USA) and used this data to create a subset of the raw electromyographic data corresponding to the 10 qualifying laps. We bandpass filtered this data using a 4th order Butterworth filter with corner frequencies of 20Hz and 500Hz. Then we used a short-time Fourier transform with a window length of 0.125s and a window overlap of 0.0625s to calculate the median frequency of the EMG power spectrum of each muscle. The median frequency data were normalized to a percentage of the maximum median frequency per muscle per test.

Statistical Analysis

We used two level mixed effects growth models with maximum likelihood estimation to analyze within-driver and between-driver change in both lap time and normalized median frequency (NMF). We chose to use individual growth curve (IGC) models instead of the more traditional repeated measures ANOVA for four reasons. First, our analysis is longitudinal; that is, it examines intra-individual change over time. Therefore, it is unlikely that each observation is truly independent. This violates the assumption of independence of observations. Second, our design is unbalanced (unequal sample sizes). Using ANOVA in these circumstances increases the risk of Type I error compared to IGC models. Third, IGC models permit the examination of both individual and group level (aggregated) curves. In contrast, repeated measures ANOVA only allows for group level analysis. Fourth, the effects of both invariant and time-variant predictors can be added to IGC models to examine associations between predictors and change in the dependent variables over time. Repeated measures ANOVA lacks such flexibility. (See Shek and Ma [2011] and Singmann and Kellen [2019] for more detailed information on the advantages of IGC models for the analysis of time series data).

Our Level 1 IGC model for lap time examined within-driver change across 10 matched pairs of laps per driver. Hence, lap times—and not drivers—are the unit of observation (what is being measured) for the lap time analysis. Similarly, our Level 1 model for the muscle activity data examined within-driver change in normalized median frequency across matched pairs of measures taken every 0.0625s for the duration of the session. The Level 2 models examined between-driver change in lap time and NMF for the four drivers across the two sessions. Hence, total sample size for the Level 2 lap time model was 75 laps (Driver 3 missed 5 laps in the session driven without the insert). Total sample size for the Level 2 NMF models was 147044 samples per muscle.

Our analytic strategy involved progressively testing unconditional linear, quadratic, and cubic trends for model fit. For the muscle activity data, this was done by grouping the seven muscles sampled into three distinct anatomical regions: the neck, the shoulder, and the trunk. Model fit was determined based on the results of Chi-square likelihood ratio tests. A quadratic trend was tested only if the linear trend was statistically significant. Likewise, a cubic trend was tested only if the linear and quadratic trends were statistically significant. This approach is consistent with the recommendations of Singer and Willet (Singer & Willett, 2003) and Field (Field, 2013) for using growth models to examine rates of change over time and identical to the approach in previous studies of steering behavior of race car drivers (Rosalie & Malone, 2018a, 2018b, 2019a, 2019b).

Time-invariant predictors were added to the Level 2 models with the best fit to investigate whether driving position (i.e., Generic and Custom) was a predictor of lap time and NMF (i.e., a fixed effect). In addition, individual differences were examined by progressively specifying random effects for the intercept, slope, and both the intercept and the slope using a heterogenous first order autoregressive or variance components covariance structure.

Again, model fit was tested using Chi-square likelihood ratio tests. For the lap time data, the intercept corresponds to lap time in the first lap and the slope to how lap time changes over the ten laps. For the NMF data, we have reported only the slope of the fitted growth curve which corresponds to fatigue resulting from muscle activity (Cifrek et al., 2000; Roy & De Luca, 1989). Muscle use, which results in fatigue, causes a downward shift in NMF which is represented by a negative slope (Cifrek et al., 2000; Phinyomark et al., 2012; Roy & De Luca, 1989). However, the model still retains the y-intercept to control for differences in initial contractile level (that is, unmodified by muscle use) between conditions (Cifrek et al., 2000; Roy & De Luca, 1989).

Results

Lap Time and Driving Position

Each driver drove ten laps in a generic seat followed by ten laps in a customized seat except for Driver 3, who completed only five laps in the generic seat because she was dissatisfied with her driving position. Drivers 1, 2, and 3, who all improved their lap time, drove a combined total of 55 laps (n = 55). Driver 4, who did not improve her lap time, drove a combined total of 20 laps (n = 20). The results of the lap time analysis are reported in Table 1.

Lap times neither changed significantly across the ten-lap session (p = 0.539), nor varied significantly across drivers ($p_{var(u0j)}$ = 0.21, $p_{var(u1j)}$ = 0.97). Drivers 1, 2 and 3 improved their lap times by an average 4.60s (SE = 1.4s) with the seat insert fitted (p = 0.003). In contrast, Driver 4 was 0.35 seconds slower with the seat insert; however, the difference was not significant (p = 0.72). The insert for Driver 4 was molded without our oversight. Given the difference in performance outcome for Drivers 1, 2 and 3 compared to Driver 4, a grouping factor (No Improvement vs. Improved) was added for subsequent analyses. Cervical and lumbar angles from the Improved and No Improvement Groups are shown in Table 2.

Table 1. Lap time in seconds for each driver with respect to driving position

Driver	1		2		3		4	
Lap	Generic	Custom	Generic	Custom	Generic	Custom	Generic	Custom
1	134.09	138.48	125.46	126.74	143.76	134.40	120.91	119.90
2	135.58	136.61	124.38	127.31	138.50	133.20	121.43	120.98
3	146.00	138.51	124.30	127.84	135.70	135.00	120.29	126.20
4	142.06	137.83	124.58	127.13	136.20	134.90	120.64	123.32
5	141.23	133.63	156.82	126.34	154.40	136.20	122.61	120.58
6	138.57	134.88	126.06	124.96		131.60	120.29	120.10
7	146.07	134.99	124.34	125.70		132.50	122.23	120.58
8	141.70	135.66	125.23	125.08		133.30	123.53	127.69
9	143.61	135.62	124.99	125.20		131.30	121.06	118.64
10	141.80	138.51	126.44	125.77		129.30	121.60	120.09
Mean	141.07	136.47	128.26	126.21	141.71	133.17	121.46	121.81
SD	3.98	1.78	10.06	1.02	7.78	2.07	1.06	2.97
Performance Outcome	Improved		Improved		Improved		No Improvement	

Table 2. Cervical and lumbar angles grouped by the effect.

	Improved		No Improvement	
	Generic	Custom	Generic	Custom
Cervical angle				
Mean	25.02	18.71	26.50	26.37
SD	7.48	6.04	1.38	0.90
Lumbar angle				
Mean	29.24	22.46	28.30	27.06
SD	5.84	8.82	0.65	0.15

Note: Angles are expressed in terms of degrees from vertical.

Neck Muscle Activity

The results of the analyses of neck muscle activity are reported in Table 3 (Appendix A) and the fitted conditional models are shown in Figure 1. A significant Group (No Improvement vs. Improved) X Condition (Generic vs. Custom) interaction, $F(1, 587814) = 1560.56$, $p < 0.001$ suggested that the effect of the seat insert on neck muscle activity differed between groups. In the No Improvement group, the seat insert was associated with significantly decreased activity of left sternocleidomastoid ($\beta = 14.70$, $p < 0.001$) and left cervical erector spinae ($\beta = 29.31$, $p < 0.001$) and significantly increased activity of right sternocleidomastoid ($\beta = -22.40$, $p < 0.001$). The effect sizes for right sternocleidomastoid ($\beta = -22.40$, $p < 0.001$) and left cervical erector spinae compared to left sternocleidomastoid suggest an increase in right lateral flexion and a decrease in right rotation in right hand corners.

In contrast, in the Improved group, the seat insert was associated with significantly decreased activity of left sternocleidomastoid ($\beta = 9.65$, $p < 0.001$), right sternocleidomastoid ($\beta = 10.66$, $p < 0.001$) and right cervical erector spinae ($\beta = 13.95$, $p < 0.001$) and increased activity of left cervical erector spinae ($\beta = -18.11$, $p < 0.001$). The effect sizes suggest an overall decrease in neck muscle use.

Results for the random effects test for individual differences in the Improved group revealed that neither intercepts (initial contractile level) nor slopes (rates of fatigue) varied between drivers for any of the four neck muscles. However, intercepts and slopes for left cervical erector spinae negatively and significantly covaried, $\mathrm{cov}\,(u_{0j}, u_{1j}) = -0.82$, $p < 0.001$ indicating that as initial contractile level increased rate of fatigue also increased. In contrast, intercepts, and slopes for right cervical erector spinae positively and significantly covaried, $\mathrm{cov}\,(u_{0j}, u_{1j}) = 0.67$, $p = 0.01$ indicating that as initial contractile level increased so did the rate of change in activity.

Shoulder Muscle Activity

The results of the analyses of shoulder muscle activity are reported in Table 4 (Appendix B) and the fitted conditional models are shown in Figure 2. A significant Group (No Improvement vs. Improved) X Condition (Generic vs. Custom) interaction, $F(1, 588166.79) = 651.08$, $p < 0.001$, suggests that the effect of the seat insert on shoulder muscle activity differed between groups. In the No Improvement group, the seat insert was associated with significantly increased activity of left anterior deltoid ($\beta = -11.08$, $p = 0.001$) and significantly decreased activities of right pectoralis major ($\beta = 19.09$, $p < 0.001$) and right anterior deltoid ($\beta = 13.49$, $p < 0.001$). The effect sizes for left anterior deltoid and right pectoralis major suggest a shift in the production of torque for right turns towards the left anterior deltoid. However, the effect sizes for right anterior deltoid and left pectoralis major ($\beta = 2.94$, $p = 0.15$) do not suggest a similar change for left hand turns largely because the latter is not significantly different.

In the Improved group, the seat insert was associated with significantly increased activity of right anterior deltoid ($\beta = -19.05$, $p < 0.001$) and right pectoralis major ($\beta = -8.74$, $p < 0.001$) which suggest that the production of steering torque shifted towards the right upper limb for both left and right turns.

Results for the random effects test for individual differences in the Improved group revealed that neither intercepts (initial contractile level) nor slopes (rates of fatigue) varied between drivers for any of the four muscles of the shoulder girdle.

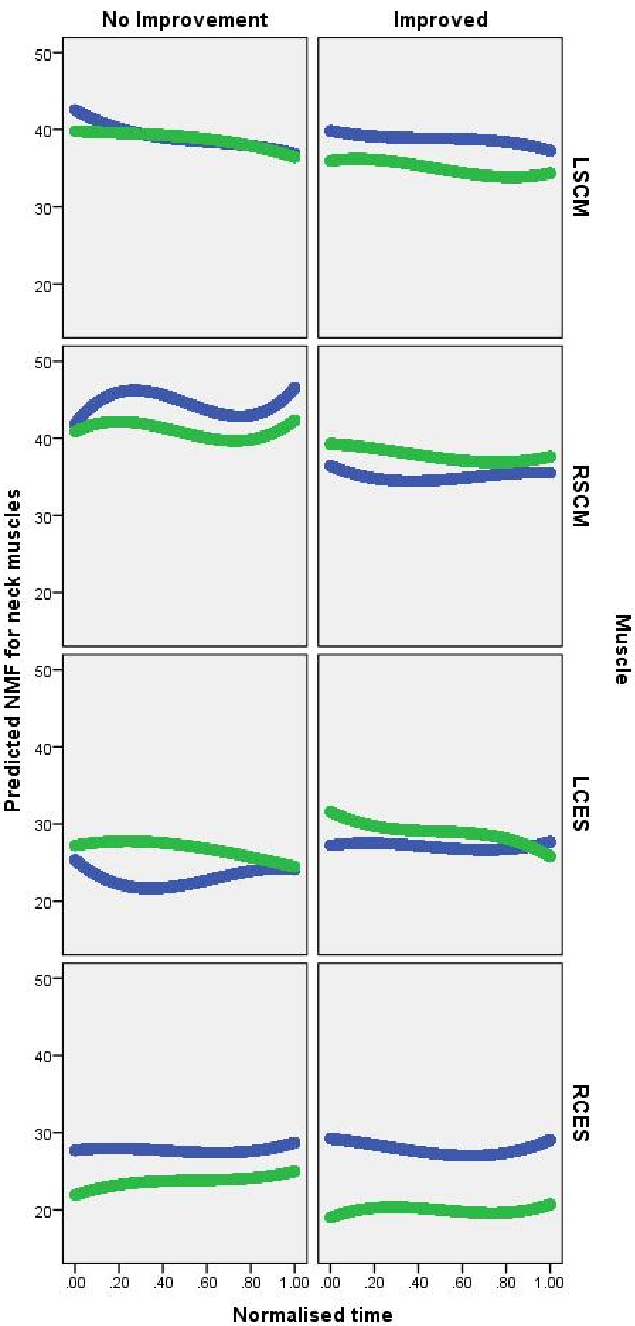

Figure 1. Neck muscle activity: fitted conditional growth curves showing the effect of driving position on normalized median frequency of left sternocleido-mastoid (LSCM), right sternocleidomastoid (RSCM), left cervical erector spinae (LCES) and right cervical erector spinae (RCES). The blue curves correspond to Generic and the green curves to Custom.

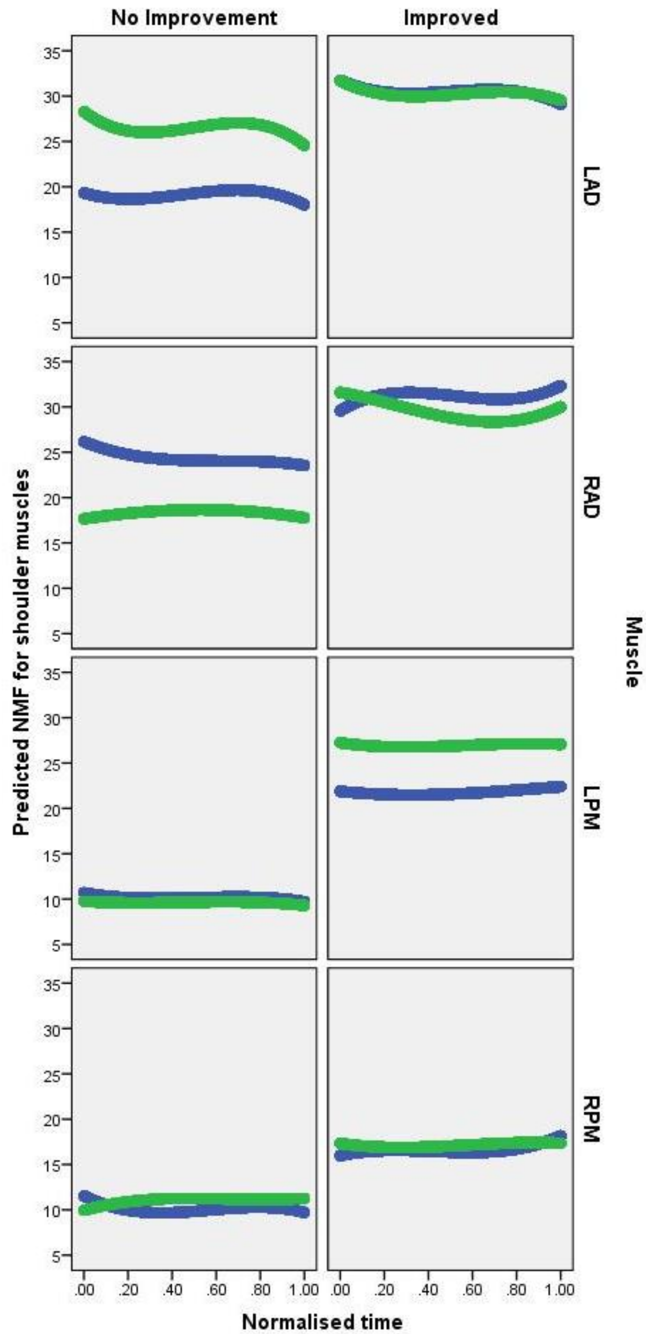

Figure 2. Shoulder muscle activity: fitted conditional growth curves showing the effect of driving position on normalized median frequency of left anterior deltoid (LAD), right anterior deltoid (RAD), left pectoralis major (LPM) and right pectoralis major (RPM). The blue curves correspond to Generic and the green curves to Custom.

Trunk Muscle Activity

The results of analyses of the trunk muscle activity are reported in Table 5 (Appendix C) and the fitted conditional models are shown in Figure 3. A significant Group (No Improvement vs. Improved) X Condition (Generic vs. Custom) interaction, $F(1, 882227.60) = 876.19$, $p < 0.001$, suggested that the effect of the seat insert on trunk muscle activity likewise differed between groups. In the No improvement group, the seat insert was associated with significantly increased activity of right lumbar erector spinae ($\beta = -11.27$, $p = 0.05$) and significantly decreased activities of left rectus abdominis ($\beta = 47.29$, $p < 0.001$), left transversus abdominis ($\beta = 22.90$, $p < 0.001$) and right transversus abdominis ($\beta = 14.09$, $p = 0.008$).

In the Improved group, the seat insert was associated with significant decreases in the activities of right lumbar erector spinae ($\beta = 6.98$, $p = 0.01$), right rectus abdominis ($\beta = 12.33$, $p < 0.001$). and right transversus abdominis ($\beta = 11.58$, $p < 0.001$) and a significant increase in the activities of left transversus abdominis ($\beta = -20.45$, *ps* < 0.001).

Results for the random effects test for individual differences in the Improved group revealed that neither intercepts (initial contractile level) nor slopes (rates of fatigue) varied between drivers for any of the six trunk muscles.

Discussion

Based on the existing literature, there are two potential explanations for why the seat insert improved the performance of three drivers, but not the fourth. One, improved lap times were only recorded for drivers whose cervical angle was decreased numerically by the seat insert. Presumably, this reduction in cervical angle resulted in the bilateral decrease in sternocleidomastoid activity. Previous research has shown that when seated in a reclined position, a bilateral decrease in the activity of sternocleidomastoid is consistent with a decrease in neck flexion (Smulders et al., 2019). However, the decrease that we measured in sternocleidomastoid activity was asymmetrical and accompanied by an asymmetrical change in

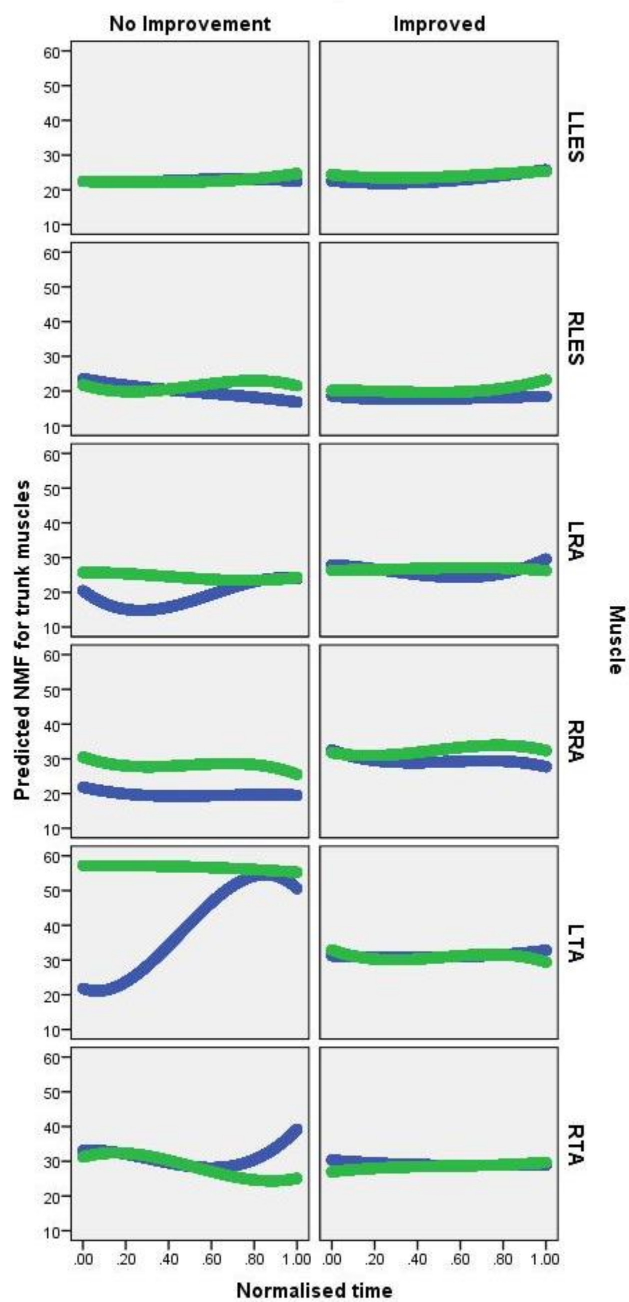

Figure 3. Trunk muscle activity: fitted conditional growth curves showing the effect of driving position on normalized median frequency (NMF) of left lumbar erector spinae (LLES), right lumbar erector spinae (RLES), left rectus abdominis (LRA), right rectus abdominis (RRA), left transversus abdominis (LTA) and right transversus abdominis (RTA). The blue curves correspond to Generic and the green curves to Custom.

activity of cervical erector spinae. This suggests that the seat insert modified the neck movement pattern of the drivers. The larger effect size for right sternocleidomastoid and the numerical increase in activation of right cervical erector spinae suggests that right lateral flexion decreased (Moroney et al., 1988).

In contrast, lap times did not improve for the driver whose seat insert did not alter cervical angle. The sternocleidomastoid activity of this driver increased on the right and decreased on the left. Most of the corners on the test track were right-hand corners. So, a forward head position in this test is likely to increase lateral flexion to the right resulting in increased activity of right sternocleidomastoid (Kim, 2015). The resulting misalignment in visual and proprioceptive information is likely to cause the driver to ‘undershoot’ right hand corners by deviating her perceived head angle to the left (Bove et al., 2001; Land & Tatler, 2001; Pettorossi & Schieppati, 2014).

An alternative explanation is that the seat insert could have increased the stability of the drivers who improved their lap times. However, the trunk muscle activities of the drivers who improved changed asymmetrically, which is not consistent with this explanation. Instead, the activity of all three muscles measured on the right side of the trunk decreased while that of left transversus abdominis increased. This probably reflects adoption of a more right arm dominant steering pattern with increased activity of right anterior deltoid and right pectoralis major. Right anterior deltoid produces positive tangential steering torque for left-hand corners and right pectoralis major produces negative tangential steering torque for right-hand corners (Pick & Cole, 2006). Trunk extensors are thought to act to oppose torque produced by the opposite upper limb while trunk flexors may act non-directionally (Hodges et al., 2000; Hodges et al., 1999; Marshall & Murphy, 2003).

The shoulder and trunk muscle activation patterns of the driver who did not improve her lap times provides further evidence that a shift in the production of steering torque is responsible for a change in trunk muscle activation. In this driver, a decrease in the activity of muscles of the right shoulder girdle and an increase in the activity of left anterior deltoid suggests that the production of steering torque, at least for right hand corners, shifted towards the production of positive tangential torque by the left upper limb. Presumably, the activities of right lumbar erector spinae and right transversus abdominis increased to balance this torque. Notably, the effect of the seat insert on the shoulder and trunk muscle activities of this driver are almost a mirror image of the changes measured in the drivers who improved their lap times. One possible explanation is that drivers who improved their laps time were in a different position relative to the steering wheel and the gear lever. This difference in position favored different strategies to achieve the same goals of steering and changing gear thereby modifying the activation patterns of shoulder and trunk muscles. The phenomenon whereby different movement strategies are used to achieve the same goal is known as motor equivalence.

This study is the fifth replication of Rosalie’s (2015) methodology published in the peer reviewed scientific literature. Two studies, this and Rosalie and Malone (2019b), have demonstrated that driver performance improves when neck lateral flexion (head tilt during cornering) is decreased. The three remaining studies all showed that neck lateral flexion increases when the demands placed on the attention of the driver are increased, either by intentionally following another car (Rosalie & Malone, 2019a) or by an unfamiliar visual distraction (Rosalie & Malone, 2018a, 2018b).

We have also used this methodology in a “clinical” setting. By measuring how the neuromuscular system of the driver responds to various challenges on track and combining these measurements with vehicle telemetry, onboard video, and accurate GPS, it is possible to predict track locations and race conditions that will that cause the driver to make an error (leading to an accident), prescribe remedial training the treat the underlying cause, and return to the driver to the track with improved performance (Rosalie, 2020).

The common finding in both research and clinician settings is that excess neck lateral flexion reduces the accuracy of visual perception in race drivers leading to a decline in performance. The reason why this occurs is complicated. Unlike road drivers who use eye movements to navigate bends (Land & Lee, 1994), race drivers navigate via head rotation (Land & Tatler, 2001). Fans of motorsport will be familiar with drivers tilting their heads (neck lateral flexion) into a corner, particularly in high *g* motorsports such as Formula 1. While this phenomenon is typically attributed to alignment of the head with centripetal (cornering) forces, head tilt is poorly correlated with centripetal force (Zikovitz & Harris, 1999). Instead, it is thought that drivers tilt into a corner to maintain a stable visual reference for the curvature of the road (Zikovitz & Harris, 1999). This automated response, known as the optokinetic cervical reflex, has also been observed in pilots of high performance aircraft when turning into and out of a bank (Coakwell et al., 2004).

There are several reasons why reflexive lateral flexion is not considered an *optimal* movement pattern for a racing driver. Tilting the head during cornering can cause conflict between the three perceptual systems that provide information for orientation: the visual system, the vestibular system, and the proprioceptive system (Fouque et al., 1999). Consider a racing driver taking a right-hand corner on a racetrack. First, the driver rotates their head to visualize the apex approximately 1s before changing steering wheel angle to rotate the car (Land & Tatler, 2001). Second, they tilt their head to the right to align with visual road tilt (Zikovitz & Harris, 1999). Consequently, the visual reference frame of the driver no longer aligns with the visual reference frame of the track (unless there is significant camber). This could cause the driver to misjudge their heading (Treffner et al., 2002).

In addition, the head tilt of the driver will cause conflict between the proprioceptive information from left sternocleidomastoid muscle which rotates the head ("I am turning right.") and right sternocleidomastoid which tilts the head to right *but also rotates the head to the left* ("I am turning left.") (Bove et al., 2001; Pettorossi & Schieppati, 2014). When drivers integrate the information from the visual and vestibular systems with the information from the proprioceptive system, they could perceive that they are understeering and overcorrect.

Despite the availability of sensory information from the visual, vestibular, and proprioceptive systems, information from each system may not be equally weighted, or indeed used at all, to inform perception in all situations. Because driving is predominantly a visual task (Zikovitz & Harris, 1999), errors stemming from erroneous proprioceptive information are more likely to occur when the demands on visual attention increase. For example, when a driver intentionally follows another car to overtake (Rosalie & Malone, 2019a). Increased demands on attention have been shown to reduce driver performance (Baldisserri et al., 2014). Therefore it is critical that the coordination patterns that drivers use optimize the accuracy of each source of information integrated into their "global-array" (Stoffregen & Bardy, 2001). As a result, when one sensory system is "occupied" (e.g., vision to avoid a collision or audition to listen to team radio) the remaining senses support accurate track navigation. We have shown repeatedly that it is possible to improve driver performance and reduce driver error by decreasing head tilt during race driving. Therefore, we recommend further close investigation of the potential of our methodology to improve driver performance by understanding how the neuromuscular system of the driver responds to various tasks on track.

Limitations and Future Directions

The main limitation of our study is the small sample size. Therefore, care should be taken in generalizing our results beyond the population of amateur female racing drivers driving closed cockpit race cars. However, small sample size is a common limitation in the scientific study of on-track driver performance which should not necessarily limit translation of research findings into performance and safety benefits. Case studies (e.g., Land & Tatler, 2001) and investigations of two or three drivers within one team (e.g., Potkanowicz et al., 2020) are the norm rather than the exception. Larger cohort studies such as the one by Carlson et al. (2014) are a rarity (*n* = 8) and virtually unheard of at elite level (e.g., Formula 1). Cost, regulations that restrict practice and the understandable unwillingness of teams and drivers to share innovations with the potential to benefit performance all contribute to limiting the number of drivers available for a given study. Ideally, our study should be replicated with a larger cohort of drivers all of whom complete three tests: a baseline, a test with a sham seat insert that does not improve driving position, and an "active" insert that does. Drivers should complete the three tests in a random order to which both researchers and drivers are blinded. This type of design would suit a single manufacturer series, such as FIA W Series, where driver performance has greater influence on the overall outcome. If the past limitations on sample size in scientific study in motorsport continue, it's important to recognize that understanding how to achieve truly exceptional performance can be realized only through researching the performance of truly exceptional individuals (e.g., Hoogkamer et al., 2019).

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Authors' Declarations

The authors declare that there are no personal or financial conflicts of interest regarding the research in this article.

The authors declare that they conducted the research reported in this article in accordance with the Ethical Principles of the Journal of Expertise.

The authors declare that they are not able to make the dataset publicly available but are able to provide it upon request.

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Appendix A. Table 3. Parameter estimates for the fixed and random effects of the seat insert on NMF of left sternocleidomastoid (LSCM), right sternocleidomastoid (RSCM), left cervical erector spinae (LCES) and right cervical erector spinae (RCES) with respect to group (Improved vs. No Improvement)

Estimates of Fixed Effects

Muscle	Group	Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
								Lower Bound	Upper Bound
LSCM	No Improvement	Time	-16.16	2.52	39170.00	-6.42	<0.001	-21.09	-11.23
		$Time^2$	22.87	5.85	39170.00	3.91	<0.001	11.40	34.33
		$Time^3$	-12.44	3.84	39170.00	-3.23	0.001	-19.97	-4.90
		Time X Condition	14.70	3.56	39170.00	4.13	<0.001	7.73	21.67
		$Time^2$ X Condition	-20.91	8.27	39170.00	-2.53	0.01	-37.11	-4.71
		$Time^3$ X Condition	8.58	5.43	39170.00	1.58	0.11	-2.07	19.23
	Improved	Time	-5.64	2.07	130.57	-2.73	0.007	-9.73	-1.55
		$Time^2$	11.84	4.43	107868.03	2.67	0.008	3.16	20.52
		$Time^3$	-8.70	2.91	107868.03	-2.99	0.003	-14.40	-2.99
		Time X Condition	9.65	2.72	107868.04	3.55	<0.001	4.33	14.98
		$Time^2$ X Condition	-30.69	6.31	107868.03	-4.86	<0.001	-43.06	-18.31
		$Time^3$ X Condition	22.01	4.15	107868.03	5.30	<0.001	13.87	30.14
RSCM	No Improvement	Time	36.92	3.25	39170.00	11.36	<0.001	30.55	43.29
		$Time^2$	-92.93	7.55	39170.00	-12.31	<0.001	-107.73	-78.13
		$Time^3$	60.72	4.96	39170.00	12.23	<0.001	50.99	70.45
		Time X Condition	-22.40	4.59	39170.00	-4.88	<0.001	-31.40	-13.40
		$Time^2$ X Condition	46.31	10.67	39170.00	4.34	<0.001	25.40	67.23
		$Time^3$ X Condition	-27.13	7.01	39170.00	-3.87	<0.001	-40.88	-13.38
	Improved	Time	-12.24	2.13	395.13	-5.76	<0.001	-16.42	-8.06
		$Time^2$	23.03	4.73	107868.08	4.87	<0.001	13.76	32.30
		$Time^3$	-11.68	3.11	107868.08	-3.76	<0.001	-17.77	-5.59
		Time X Condition	10.66	2.90	107868.09	3.68	<0.001	4.98	16.35
		$Time^2$ X Condition	-31.25	6.74	107868.08	-4.64	<0.001	-44.46	-18.04
		$Time^3$ X Condition	19.87	4.43	107868.08	4.48	<0.001	11.18	28.55

LCES	No Improvement	Time	-24.38	3.19	39170.00	-7.63	<0.001	-30.64	-18.12
		$Time^2$	48.57	7.42	39170.00	6.54	<0.001	34.02	63.12
		$Time^3$	-25.43	4.88	39170.00	-5.21	<0.001	-35.00	-15.87
		Time X Condition	29.31	4.51	39170.00	6.49	<0.001	20.46	38.16
		$Time^2$ X Condition	-60.49	10.49	39170.00	-5.77	<0.001	-81.05	-39.93
		$Time^3$ X Condition	29.68	6.90	39170.00	4.30	<0.001	16.16	43.19
	Improved	Time	4.58	2.78	23.46	1.65	0.11	-1.17	10.34
		$Time^2$	-14.74	5.20	107867.99	-2.83	0.005	-24.93	-4.55
		$Time^3$	11.23	3.42	107867.99	3.28	0.001	4.53	17.93
		Time X Condition	-18.11	3.19	107867.99	-5.68	<0.001	-24.36	-11.86
		$Time^2$ X Condition	42.29	7.41	107867.99	5.71	<0.001	27.76	56.81
		$Time^3$ X Condition	-30.46	4.87	107867.99	-6.25	<0.001	-40.01	-20.91
RCES	No Improvement	Time	3.32	2.79	39170.00	1.19	0.23	-2.15	8.80
		$Time^2$	-12.45	6.49	39170.00	-1.92	0.05	-25.16	0.26
		$Time^3$	10.13	4.26	39170.00	2.38	0.02	1.78	18.49
		Time X Condition	6.23	3.94	39170.00	1.58	0.11	-1.50	13.96
		$Time^2$ X Condition	-4.13	9.17	39170.00	-0.45	0.65	-22.10	13.84
		$Time^3$ X Condition	-0.04	6.03	39170.00	-0.01	1.00	-11.85	11.78
	Improved	Time	-2.11	2.51	15.32	-0.84	0.41	-7.44	3.23
		$Time^2$	-7.16	4.37	107868.04	-1.64	0.10	-15.72	1.39
		$Time^3$	9.87	2.87	107868.04	3.44	0.001	4.25	15.50
		Time X Condition	13.95	2.68	107868.05	5.21	<0.001	8.70	19.20
		$Time^2$ X Condition	-19.60	6.22	107868.04	-3.15	0.002	-31.80	-7.40
		$Time^3$ X Condition	7.59	4.09	107868.04	1.85	0.06	-0.43	15.61

Estimates of the Random Effects

Muscle	Group	Parameter	Estimate	Std. Error	Wald Z	Sig.	95% Confidence Interval	
							Lower Bound	Upper Bound
LSCM	Improved	Variance in Intercepts	30.48	24.90	1.22	0.22	6.14	151.20
		Variance in Time		1.58	1.20	0.23	0.37	9.75
		Covariance	1.89		0.48	0.63	-0.71	0.89
RSCM	Improved	Variance in Intercepts	64.80	52.94	1.22	0.22	13.07	321.30
		Variance in Time	1.13	0.96	1.17	0.24	0.21	6.02
		Covariance	0.09	0.59	0.16	0.87	-0.79	0.85
LCES	Improved	Variance in Intercepts	8.40	6.87	1.22	0.22	1.69	41.77
		Variance in Time	8.24	6.79	1.21	0.23	1.64	41.46
		Covariance	-0.82	0.19	-4.40	<0.001	-0.98	-0.03
RCES	Improved	Variance in Intercepts	1.81	1.49	1.21	0.23	0.36	9.10
		Variance in Time	8.28	6.82	1.21	0.22	1.65	41.61
		Covariance	0.67	0.32	2.09	0.04	-0.33	0.96

Appendix B. Table 4. Parameter estimates for the fixed and random effects of the seat insert on NMF of left anterior deltoid (LAD), right anterior deltoid (RAD), left pectoralis major (LPM), and right pectoralis major (RAD) with respect to group (Improved vs. No Improvement).

Estimates of Fixed Effects

Muscle	Group	Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
								Lower Bound	Upper Bound
LAD	No Improvement	Time	-7.31	2.35	39170.00	-3.12	0.002	-11.91	-2.71
		$Time^2$	22.92	5.45	39170.00	4.20	<0.001	12.24	33.61
		$Time^3$	-16.92	3.58	39170.00	-4.72	<0.001	-23.94	-9.89
		Time X Condition	-11.08	3.32	39170.00	-3.34	0.001	-17.58	-4.59
		$Time^2$ X Condition	21.97	7.70	39170.00	2.85	0.004	6.87	37.07
		$Time^3$ X Condition	-13.31	5.06	39170.00	-2.63	0.01	-23.23	-3.38
	Improved	Time	-10.99	1.67	2645.74	-6.57	<0.001	-14.27	-7.71
		$Time^2$	25.48	3.83	107867.81	6.65	<0.001	17.97	32.99
		$Time^3$	-17.20	2.52	107867.81	-6.83	<0.001	-22.14	-12.26
		Time X Condition	-1.05	2.35	107867.83	-0.45	0.65	-5.66	3.55
		$Time^2$ X Condition	0.16	5.46	107867.81	0.03	0.98	-10.55	10.87
		$Time^3$ X Condition	1.34	3.59	107867.81	0.37	0.71	-5.70	8.38
RAD	No Improvement	Time	-9.87	2.29	39170.00	-4.30	<0.001	-14.36	-5.37
		$Time^2$	15.61	5.33	39170.00	2.93	0.003	5.17	26.05
		$Time^3$	-8.36	3.50	39170.00	-2.39	0.02	-15.22	-1.49
		Time X Condition	13.49	3.24	39170.00	4.16	<0.001	7.14	19.84
		$Time^2$ X Condition	-18.67	7.53	39170.00	-2.48	0.01	-33.43	-3.91
		$Time^3$ X Condition	7.90	4.95	39170.00	1.60	0.11	-1.80	17.60
	Improved	Time	15.06	1.84	609.95	8.19	<0.001	11.45	18.67
		$Time^2$	-34.05	4.13	107866.35	-8.24	<0.001	-42.15	-25.96
		$Time^3$	21.90	2.72	107866.35	8.06	<0.001	16.57	27.22
		Time X Condition	-19.05	2.53	107866.38	-7.52	<0.001	-24.02	-14.09
		$Time^2$ X Condition	25.69	5.89	107866.35	4.36	<0.001	14.15	37.23

		$Time^3$ X Condition	-10.97	3.87	107866.35	-2.83	0.005	-18.55	-3.38
LPM	No Improvement	Time	-4.46	1.43	39170.00	-3.11	0.002	-7.27	-1.65
		$Time^2$	9.98	3.33	39170.00	3.00	0.003	3.45	16.51
		$Time^3$	-6.51	2.19	39170.00	-2.97	0.003	-10.81	-2.22
		Time X Condition	2.94	2.03	39170.00	1.45	0.15	-1.03	6.91
		$Time^2$ X Condition	-5.69	4.71	39170.00	-1.21	0.23	-14.92	3.53
		$Time^3$ X Condition	3.30	3.09	39170.00	1.07	0.29	-2.76	9.37
	Improved	Time	-2.70	2.03	3001.87	-1.33	0.18	-6.68	1.27
		$Time^2$	4.89	4.65	107867.99	1.05	0.29	-4.22	14.00
		$Time^3$	-1.76	3.06	107867.99	-0.58	0.56	-7.75	4.23
		Time X Condition	-0.84	2.85	107868.01	-0.30	0.77	-6.43	4.74
		$Time^2$ X Condition	2.84	6.62	107867.99	0.43	0.67	-10.15	15.82
		$Time^3$ X Condition	-2.68	4.35	107867.99	-0.62	0.54	-11.22	5.85
RPM	No Improvement	Time	-12.35	1.79	39170.00	-6.88	<0.001	-15.87	-8.84
		$Time^2$	25.26	4.17	39170.00	6.06	<0.001	17.08	33.44
		$Time^3$	-14.73	2.74	39170.00	-5.37	<0.001	-20.10	-9.35
		Time X Condition	19.09	2.54	39170.00	7.52	<0.001	14.12	24.06
		$Time^2$ X Condition	-36.39	5.89	39170.00	-6.17	<0.001	-47.94	-24.84
		$Time^3$ X Condition	20.45	3.87	39170.00	5.28	<0.001	12.85	28.04
	Improved	Time	5.37	1.49	587.65	3.61	<0.001	2.45	8.29
		$Time^2$	-14.84	3.34	107867.71	-4.44	<0.001	-21.39	-8.29
		$Time^3$	11.82	2.20	107867.71	5.38	<0.001	7.51	16.12
		Time X Condition	-8.74	2.05	107867.72	-4.27	<0.001	-12.76	-4.72
		$Time^2$ X Condition	23.22	4.76	107867.71	4.88	<0.001	13.89	32.55
		$Time^3$ X Condition	-16.63	3.13	107867.71	-5.31	<0.001	-22.77	-10.50

Estimates of the Random Effects

Muscle	Group	Parameter	Estimate	Std. Error	Wald Z	Sig.	95% Confidence Interval	
							Lower Bound	Upper Bound
LAD	Improved	Variance in Intercepts	11.284	9.226	1.223	0.221	2.272	56.034
		Variance in Time	0.232	0.228	1.020	0.308	0.034	1.586
RAD	Improved	Variance in Intercepts	0.088	0.090	0.979	0.328	0.012	0.654
		Variance in Time	0.648	0.578	1.120	0.263	0.113	3.726
LPM	Improved	Variance in Intercepts	11.886	9.724	1.222	0.222	2.392	59.071
		Variance in Time	0.328	0.313	1.046	0.296	0.050	2.134
RPM	Improved	Variance in Intercepts	0.195	0.167	1.164	0.244	0.036	1.050
		Variance in Time	0.435	0.386	1.126	0.260	0.076	2.479

Appendix C. Table 5. Parameter estimates for the fixed and random effects of the seat insert on NMF of left lumbar erector spinae (LLES), right lumbar erector spinae (RLES), left rectus abdominis (LRA), right rectus abdominis (RRA), left transversus abdominis (LTA) and right transversus abdominis (RTA) with respect to group (Improved vs. No Improvement)

Estimates of Fixed Effects

Muscle	Group	Parameter	Estimate	Std. Error	df	t	Sig.	95% Confidence Interval	
								Lower Bound	Upper Bound
LLES	No Improvement	Time	-3.29	2.66	39170.00	-1.24	0.22	-8.49	1.92
		$Time^2$	12.98	6.17	39170.00	2.10	0.04	0.88	25.07
		$Time^3$	-9.78	4.06	39170.00	-2.41	0.02	-17.73	-1.83
		Time X Condition	2.58	3.75	39170.00	0.69	0.49	-4.77	9.94
		$Time^2$ X Condition	-14.24	8.72	39170.00	-1.63	0.10	-31.33	2.85
		$Time^3$ X Condition	14.04	5.73	39170.00	2.45	0.01	2.81	25.27
LLES	Improved	Time	-5.58	2.38	17.49	-2.35	0.03	-10.58	-0.58
		$Time^2$	11.66	4.24	107867.98	2.75	0.006	3.36	19.97
		$Time^3$	-3.10	2.79	107867.98	-1.11	0.27	-8.56	2.36
		Time X Condition	-1.49	2.60	107867.98	-0.57	0.57	-6.58	3.61
		$Time^2$ X Condition	2.30	6.04	107867.98	0.38	0.70	-9.54	14.13
		$Time^3$ X Condition	-3.09	3.97	107867.98	-0.78	0.44	-10.87	4.69
RLES	No Improvement	Time	-11.27	2.73	39170.00	-4.13	<0.001	-16.63	-5.92
		$Time^2$	10.49	6.35	39170.00	1.65	0.10	-1.96	22.94
		$Time^3$	-5.96	4.18	39170.00	-1.43	0.15	-14.14	2.22
		Time X Condition	-7.56	3.86	39170.00	-1.96	0.05	-15.13	0.01
		$Time^2$ X Condition	42.49	8.98	39170.00	4.73	<0.001	24.90	60.09
		$Time^3$ X Condition	-28.47	5.90	39170.00	-4.83	<0.001	-40.04	-16.91
RLES	Improved	Time	-6.74	1.67	64.08	-4.03	<0.001	-10.08	-3.40
		$Time^2$	12.84	3.45	107868.00	3.72	<0.001	6.08	19.59
		$Time^3$	-6.37	2.27	107868.00	-2.81	0.005	-10.81	-1.93
		Time X Condition	6.98	2.11	107868.00	3.30	0.001	2.83	11.12
		$Time^2$ X Condition	-21.35	4.91	107868.00	-4.35	<0.001	-30.98	-11.72

		$Time^3$ X Condition	17.69	3.23	107868.00	5.48	<0.001	11.36	24.02
LRA	No Improvement	Time	-46.86	3.90	39170.00	-12.02	<0.001	-54.51	-39.22
		$Time^2$	112.33	9.06	39170.00	12.40	<0.001	94.57	130.09
		$Time^3$	-62.09	5.96	39170.00	-10.42	<0.001	-73.77	-50.41
		Time X Condition	47.29	5.51	39170.00	8.58	<0.001	36.49	58.09
		$Time^2$ X Condition	-125.20	12.81	39170.00	-9.78	0.001	-150.30	-100.10
		$Time^3$ X Condition	73.07	8.42	39170.00	8.68	<0.001	56.57	89.57
LRA	Improved	Time	-3.91	2.27	412.03	-1.73	0.09	-8.36	0.54
		$Time^2$	-17.52	5.05	107868.08	-3.47	0.001	-27.41	-7.63
		$Time^3$	22.96	3.32	107868.08	6.92	<0.001	16.46	29.46
		Time X Condition	3.34	3.09	107868.09	1.08	0.28	-2.72	9.41
		$Time^2$ X Condition	24.03	7.19	107868.08	3.34	0.001	9.94	38.13
		$Time^3$ X Condition	-29.19	4.73	107868.08	-6.17	<0.001	-38.45	-19.92
RRA	No Improvement	Time	-13.97	4.02	39170.00	-3.48	0.001	-21.84	-6.10
		$Time^2$	23.73	9.33	39170.00	2.54	0.01	5.44	42.02
		$Time^3$	-12.10	6.13	39170.00	-1.97	0.05	-24.12	-0.08
		Time X Condition	-7.79	5.67	39170.00	-1.37	0.17	-18.91	3.34
		$Time^2$ X Condition	27.95	13.19	39170.00	2.12	0.03	2.10	53.79
		$Time^3$ X Condition	-22.84	8.67	39170.00	-2.63	0.008	-39.83	-5.85
RRA	Improved	Time	-23.75	3.21	9.68	-7.39	<0.001	-30.94	-16.56
		$Time^2$	45.97	4.98	107867.99	9.23	<0.001	36.21	55.73
		$Time^3$	-27.84	3.27	107867.99	-8.50	<0.001	-34.25	-21.42
		Time X Condition	12.33	3.06	107867.99	4.04	<0.001	6.34	18.32
		$Time^2$ X Condition	-8.13	7.10	107867.99	-1.15	0.25	-22.05	5.79
		$Time^3$ X Condition	1.20	4.67	107867.99	0.26	0.80	-7.95	10.35
LTA	No Improvement	Time	-22.79	3.58	39170.00	-6.37	<0.001	-29.80	-15.78
		$Time^2$	189.50	8.31	39170.00	22.80	<0.001	173.20	205.79
		$Time^3$	-137.95	5.46	39170.00	-25.24	<0.001	-148.66	-127.24

		Time X Condition	22.90	5.06	39170.00	4.53	<0.001	12.99	32.81
		$Time^2$ X Condition	-190.88	11.75	39170.00	-16.25	<0.001	-213.91	-167.86
		$Time^3$ X Condition	137.28	7.72	39170.00	17.78	<0.001	122.14	152.41
LTA	Improved	Time	-2.10	2.38	27.30	-0.88	0.39	-6.98	2.78
		$Time^2$	0.82	4.53	107867.95	0.18	0.86	-8.05	9.70
		$Time^3$	2.59	2.98	107867.95	0.87	0.38	-3.24	8.43
		Time X Condition	-20.45	2.78	107867.95	-7.36	<0.001	-25.89	-15.00
		$Time^2$ X Condition	52.31	6.46	107867.95	8.10	<0.001	39.65	64.96
		$Time^3$ X Condition	-36.89	4.24	107867.95	-8.69	<0.001	-45.21	-28.57
RTA	No Improvement	Time	2.74	3.76	39170.00	0.73	0.47	-4.63	10.11
		$Time^2$	-51.17	8.73	39170.00	-5.86	<0.001	-68.29	-34.06
		$Time^3$	54.51	5.74	39170.00	9.49	<0.001	43.26	65.77
		Time X Condition	14.09	5.31	39170.00	2.65	0.008	3.68	24.50
		$Time^2$ X Condition	-13.47	12.34	39170.00	-1.09	0.28	-37.66	10.73
		$Time^3$ X Condition	-12.83	8.11	39170.00	-1.58	0.11	-28.73	3.07
RTA	Improved	Time	-5.18	1.90	107.40	-2.73	0.007	-8.94	-1.42
		$Time^2$	4.70	4.04	107867.84	1.17	0.24	-3.21	12.62
		$Time^3$	-1.21	2.65	107867.84	-0.46	0.65	-6.41	3.99
		Time X Condition	11.58	2.48	107867.85	4.68	<0.001	6.73	16.43
		$Time^2$ X Condition	-14.96	5.75	107867.84	-2.60	0.01	-26.23	-3.68
		$Time^3$ X Condition	7.47	3.78	107867.84	1.98	0.05	0.06	14.88

Estimates of the Random Effects

Muscle	Group	Parameter	Estimate	Std. Error	Wald Z	Sig.	95% Confidence Interval	
							Lower Bound	Upper Bound
LLES	Improved	Variance in Intercepts	20.99	17.15	1.22	0.22	4.23	104.15
		Variance in Time	6.97	5.73	1.22	0.22	1.39	34.92
RLES	Improved	Variance in Intercepts	13.61	11.13	1.22	0.22	2.74	67.57
		Variance in Time	1.78	1.48	1.20	0.23	0.35	9.08
LRA	Improved	Variance in Intercepts	31.07	25.40	1.22	0.22	6.26	154.19
		Variance in Time	1.25	1.07	1.17	0.24	0.23	6.70
RRA	Improved	Variance in Intercepts	74.09	60.52	1.22	0.22	14.94	367.35
		Variance in Time	17.16	14.07	1.22	0.22	3.44	85.59
LTA	Improved	Variance in Intercepts	5.73	4.70	1.22	0.22	1.15	28.61
		Variance in Time	5.57	4.59	1.21	0.23	1.10	28.04
RTA	Improved	Variance in Intercepts	4.62	3.79	1.22	0.22	0.93	23.03
		Variance in Time	1.75	1.47	1.19	0.24	0.33	9.13