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Development of a Lower Extremity Mobility Assessment Methodology for Motor Vehicle Operation and Initial Validation

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DEVELOPMENT OF A LOWER EXTREMITY
MOBILITY ASSESSMENT METHODOLOGY FOR
MOTOR VEHICLE OPERATION AND INITIAL
VALIDATION

A Thesis
Presented to
the Graduate School of
Clemson University

In Partial Fulfillment
of the Requirements for the Degree
Master of Science
in Bioengineering

by
Justin Arnosky
May 2012

Accepted by:
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ABSTRACT

Limited quantifiable data exists on lower extremity mobility and function during driving. To date, the most appropriate existing measures of successful driving function are assessed by a driving rehabilitation specialist during an on-road evaluation.

Establishing the kinematic chain- or the order and magnitude in which joints are moved- during driving may prove to be a useful tool in lower extremity function assessment in drivers. To this end, a study was conducted instrumenting both the left and right legs of healthy licensed male drivers (18-26 years old) with a system of angle measuring goniometers (Biometrics, Ltd.) in a driving simulator (DriveSafety CDS-250). The motions across the hip, knee and AFC joints were measured during active driving simulator scenarios, performing pedal tasks with both the right and left leg. Subjects completed 3 trials for each leg in which they were required to respond to braking tasks and peripheral queuing, and comparisons between left versus right leg driving over time were conducted for measuring brake response time, return to gas movement time, and joint angle minimums, maximums, and ranges of motion. Kinematic chain joint angles were also correlated against each other so as to yield a slope and strength of correlation, allowing the development of a numerical assessment of the kinematic chain.

Results of this work indicate that left leg driving requires characteristically different kinematic chain in lower extremity motions, primarily with respect to the altered use of AFC inversion/eversion. Left limb correlation values were found, in general, to have a higher value, indicating a greater degree of repeatable gross motor movement. Right leg motions showed a greater range of fine motor control, which could be

characteristic of dominant leg driving in general. Similar movement patterns were found in both phases of pedal transition, both the brake application and the return from brake to gas. This study showed that the distinctive motions seen in right versus left-footed driving can indeed be characterized by goniometric application. Further studies should explore the effects of left leg driver training in a longitudinal manner, testing this driving task over the period of several weeks. If these future studies show a development and improvement of left leg driver performance, patients undergoing right leg orthopedic procedures could be taught to drive effectively with the left leg during rehabilitation for extended periods of time, thereby allowing those patients to maintain their independence.

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INTRODUCTION

The standards for license suspension and termination in the United States are currently being debated and met with great opposition as driving is a benchmark of independence for Americans since license attainment for many at the age of 16. Response time, movement time, and cognitive awareness are considered critical driving performance variables, but are creating regulatory disagreement for the assessment of elderly drivers. The number fatal accidents involving drivers over the age of 70 is no different than any other age bracket¹. However, when the number of accidents per mile is considered, there is a significant increase in the proportion of accidents involving those drivers over the age of 70, and a second marked increase in drivers over 80¹. This statistic becomes more alarming with the aging population trend seen in the United States. Currently, 30 million of the 38 million individuals over the age of 65 are licensed drivers², and estimates project that by the year 2020, there will be more than 40 million licensed drivers over the age of 65 in the U.S.³.

The coordinated and deliberate movement of the lower extremity motion during driving is of particular importance during driving, as brake reaction time is a primary variable in crash prevention. Being able to characterize lower extremity driver motion could aid in the understanding of why some drivers are more at risk for certain kinds of accidents. Although left leg driving is rare, drivers have been known to drive with two feet, operating the gas and brake pedals independently in automatic transmission vehicles. Altered leg use during driving could result from persons recovering from broken feet or legs, osteoarthritic joints, or other orthopaedic procedures. By quantifying

left and right leg driving, a protocol of driver practice may allow for patients who will be undergoing surgical treatment to train in left-footed driving in a preparatory phase of surgery. This work seeks to develop a method to assess motion of the lower leg during a driving simulator scenario, and to then use this method to compare the lower extremity motion of a controlled subject cohort during a set of left vs. right leg driving tasks.

*****Note from the authors:** The research study performed did not utilize any adaptive equipment currently available. The practicality of the study was viewed by the authors as drivers being capable of using cars as issued by the manufacturer, with no modifications made to the vehicle. Adaptive equipment is currently being employed by individuals who have undergone right lower leg amputations, placing the gas pedal on the left side of the pedal well. This technology is currently under scrutiny because many complications with driving are associated with it, and further work needs to be done to establish if this is in fact a viable direction for the technology to go.

Current Methods of Driver Performance Assessment

Although the use of lower leg movement to assess driving performance measures has not been previously conducted in the literature, specific vehicle tests have been previously developed to assess individuals who have mild but potentially dangerous cognitive impairments, and these tests have been determined to be reasonably reliable and valid⁴. An off-road set of assessments lasting 60-90 minutes, when combined with the 40-60 minute on-road driving evaluation, is referred to as a comprehensive driving evaluation (CDE). The off-road portion usually includes standardized assessments that establish physical, visual, and cognitive abilities in a non-driving environment⁵.

Physicians, as well as optometrists and family members of these questionable drivers, are often given the task of referring drivers to be assessed in the CDE. However, the activities of daily living (and tasks found in the battery of off-road tests) in which physicians and family members see the senior citizen perform has been found to have no correlation with the individual's ability to pass a driving test ⁶. Additionally, conventional vision tests don't do an adequate job of testing the individual's ability to drive. The crucial aspects of driving- especially at night- are functional field of view and glare recovery. Current assessments of the field of view are found to be only "35 percent sensitive and to be inadequate for detecting visual field defects." ⁶.

Limited quantifiable data exists on the movement of automobile drivers. While there are several studies exploring the effect cognitive deficits have on driving, data relating specifically to the peripheral nervous system and actual movement of the lower extremity is severely lacking. It is well researched and known that reaction time decreases with an increased age ⁷, and this decreased reaction time is likely one of the contributing factors to an increased incidence of accidents involving elderly drivers. According to a study performed by Warshawsky-Livne et al. in 2002, reaction time is the duration between stimulus and an adequate reaction (e.g. braking). Response time is defined as the duration between the onset of braking stimulus to the application of the brake pedal. The reason it is defined as "response time" and not "reaction time" is because the driver must make a decision to complete the task- they may either depress the gas or brake pedal. Response time is being explored in the current study because the

brake response time is the most complete measure of the braking motion, showing the total time elapsed from a braking stimulus to the correct response.



Figure 1: Representation of the experimental setup used to test EMG activity in the quadriceps muscle during MMT.

Manual Muscle Testing (MMT) is the method which rehabilitative specialists use to assess an individual's ability to perform a task. This can include grip tests, isometric muscle contraction tests, and in a driving simulator scenario, MMT could assess a driver's leg strength and potentially give some insight to their ability to operate a car's brake and gas pedals. The peak contractions of muscle groups such as the knee extensors and ankle/foot complex (AFC) plantar flexors could prove to be a valuable variable for responses in emergency situations. This type of assessment may include dynamometers (as with grip testing) ⁸ and electromyography (EMG) sensors (when testing leg performance, figure 1) ⁹. MMT has become a very valuable method of testing a potential driver's ability to move their foot from brake to gas and back. The most effective method of testing the muscle ability is to find the Maximum Voluntary Isometric Contraction

(MVIC) of a muscle⁹. This is done by flexing the muscle in question as hard as possible for a sustained time. For instance, by extending the knee and clenching the thigh, the quadriceps MVIC can be found⁹. In the instance of driving, finding the MVIC's for the Tibialis Anterior, Peroneal muscles, and many other major muscle groups would be crucial for a rehabilitative specialist's assessment of a patient's ability to return to drive.

It is important to note, however, that the study performed by the authors of this paper did not include any aspects of MMT or other current performance variables such as visual and cognitive tests. The issue of MMT is addressed here simply because it is a current method of testing, and is found to be an accurate method of testing the muscles involved in driving, and is a way to screen if individuals are, in muscular terms, capable of driving. Other neurological qualifiers for driving, such as visual acuity and cognitive tests, were also not addressed in the scope of this work. However, these factors were minimized through the use of a narrowly defined set of patients to assess the gross movements a driver performs while operating a driving simulator.

Brake Reaction Timers

Very seldom do drivers not scan the environment and focus solely on the brake lights of the car in front of them. Even when this is the case, a whole new set of dangers beset the driver who doesn't constantly scan for potential hazards. Decision making skills, cognitive awareness, and quick reaction times were all incorporated in this study- something that has not been done to this level of sophistication in previous studies. Previous studies, as explored in this section, focus solely on finding brake reaction times in large part without outside distractions. The most crucial part of driving safely is

responding to a necessary or emergency situation in a timely manner. All the essential qualities of each of the brake reaction timers have been placed in a spreadsheet for simple comparison (table 1). Several studies have been performed to simulate driving scenarios and record brake response times. In a study performed by Wright et. al., surgical groin hernia patients were tested pre-operatively and 1, 3 and 6 days post-operatively¹⁰.

Reactions for both hand and foot responses were recorded, and compared to pre-operative speeds. The testing mechanism for this study was a black computer screen, which turned red at random intervals to signal a stop. Foot pedal reaction time was measured as the time it takes to move from the gas to the brake pedal¹⁰. While the design of this study was appropriate, it wasn't ideal. The color cue for a brake was accurate to what a driver may see, but the study neglected to make any consideration for outside distractions.

Typical driving has several distraction sources, and this study didn't recreate any of those distractions, whether they be from sound, peripheral vision, or otherwise. This study also didn't specify at what point a brake response was triggered, whether it be 1%, 5%, or full pedal depression. An adequate brake response is determined by the situation in which a braking action is required- this could be as varied as a vehicle stopping short or a light 300 yards ahead changing from green to red. This particular study found brake reaction times pre and postoperatively. A variable that could provide some great insight to the effect that hernia repairs may have would be how the motion changes in driving as a result of the surgery. A painful hernia would likely alter the movement patterns of the driver, and seeing how this movement changes through the recovery process would provide great incites to the characteristics of the kinematic chain.

Warshawski-Livne et. al. (2001) broke up braking time into two subsets, including reaction time (RT) and brake-movement time (MT)⁷. This study used a driving simulator developed by Baran Advanced Technologies mimicking a Volkswagen Passat. The drivers of varying age and gender “followed” a cut-out of a car placed 3 meters in front of the simulator cockpit, and responded to brake lights when they lit up. Digital sensors were placed on both the gas and brake pedal, each measuring different aspects of this braking time.

Simulator Quality/Reaction Timer	Vericom Stationary Reaction Timer (Dalury)	RT-2S Brake Reaction Timer (Dickerson)	Monash University Reaction Timer (Hau/Nguyen)	Baran Advanced Technologies Volkswagen Passat (Warshawsky-Livne)	Hernia patient reaction timer (Wright)	Brake Reaction Timer Model 3548 (Gotlin)	Doron Driving Simulator System (Kantor)	DriveSafety CDS-250 (Brooks)
Foot Pedal Reaction	X	X	X	X	X	X		X
Computer Based	X	X	X	X	X	X	X	X
Inexpensive	X	X				X		
Peripheral Distractions							X	X
Visual Cue to Stop	X	X	X	X	X	X	X	X
Realistic pedal well				X	X			X
Posturally accurate			X	X	X	X	X	X
Brake Light Stimulus				X			X	X
Red Light Brake Stimulus	X	X	X		X	X		
Realistic Steering Wheel			X	X	X	N/A	X	X

Table 1: Qualities of brake reaction timers as reported by authors. An ‘X’ represents the brake reaction timer having the corresponding quality.

The sensor on the gas pedal measured the reaction time- the time between brake light stimulus to gas pedal release⁷. The sensor on the brake pedal measured the movement time of the driver to get their foot from the gas to the brake pedal with an accuracy of 0.01 seconds. The authors found that there were no statistically significant

differences between the genders, but reaction time increased with age. One deficiency of the experimental setup was the recording of the reaction time. Because the gas pedal depression was gradual and not simply a trigger, the reaction time may be tainted by the driver's foot speed of release, and simply touching the brake is not enough to really mark an adequate braking response, and therefore a final braking time was not truly measured¹¹. The driving simulator and experimental environment was satisfactory, with the exception of a lack of outside distractions, as there would be in a real-world driving situation. With these variables being collected, the difference in reaction and movement times can easily be compared between the age segments and genders. Further analysis focusing on the motion of the drivers could show the kinematic differences through the genders and detect possible sources of the delayed reaction time, should it be movement-related in nature.

Hau et.al. (2000) and Nguyen et. al. (2000) had a similar experimental setup designed and constructed by the Department of Electrical and Computer Systems Engineering of Monash University (Australia), in which they used an actual steering wheel and a bare-bones representation of a vehicle cockpit^{12, 13}. Again, the goal of the study was to assess brake reaction times, though this time testing before and after knee arthroscopies and ligamentous repairs. So while actual parts of a car were used, the simulator still seems to be lacking as far as a realistic situation, as with the other studies discussed in this paper. Similar to the paper by Wright et. al. (1999), analyzing the kinematic chain pre- and post-operatively could shed some light to how the driver changes their movement patterns as a result of surgical repair.

The Vericom stationary reaction timer was used by Dalury et. al. (2010) in an effort to test the reaction time of patients returning from a right-leg total knee arthroplasty (TKA) in the hopes of verifying when they may return to driving, tested at 4, 6, and 8 weeks post-operation¹⁴. This reaction timer composes a desk-mounted driving wheel, two pedals at the floor, and a computer screen which, when the gas pedal was depressed, showed the virtual simulation car moving down a road. When a red light showed at the side of the screen, the driver would move their foot from the gas to the brake, and the reaction time was recorded. With the driver moving through a virtual reality environment, some of the concerns of a lack of environmental distractions were eliminated, though other problems persisted in this scenario. Based on figures shown in the publication, the simulator inadequately duplicated a driver's seat, but was likely very cheap, which was why the Vericom stationary reaction timer was used in many offices to assess a person's ability to drive. The steering wheel appeared to be too small when compared to an actual steering wheel, and the posture of the driver didn't seem accurate to that of a real-world driver. Altering the posture of a driver will change the order in which and in what proportion each of the muscle groups operate, and doesn't accurately paint a picture of a brake reaction. Trying to mimic actual driving posture may prove to have a significant impact in brake reaction time and driving performance in general.

A similar study was performed by Dickerson et. al. in 2008, which had some of the same costs and benefits. They used the RT-2S brake reaction timer produced by Advanced Therapy Product, Inc., which is advertised as "simple, lightweight...easy to carry, set up, and use." However, this simplicity was also the study's downfall for real-

world scenarios of driving¹⁵. Like many other systems used (Brake Reaction Timer Model 3548¹⁶), it adequately measured reaction time, but neglected many of the other issues faced by drivers. These simple, lightweight reaction timers have a major fault in not addressing the posture seen in a typical driver's seat. The disregard for posture and the kinematic chain in general leaves a significant amount of data unexplored.

Perhaps one of the most realistic driving simulators was used in a 2004 study performed by Kantor et. al. in which they used the Doron Driving Simulator System. This system included "a typical car seat, dashboard, steering wheel, and gear shift" giving the most realistic feel of an actual driver's seat¹⁷. This study measured reaction time in a manner that eliminated the guess work of the pedals- because there were no pedals, and reaction time was measured as a button press on the gear shift¹⁷. So while this study shows to be appropriate in recreating the driving environment, other testing needs to be performed to assess the muscular performance to assure that the driver is capable of operating the foot pedals. Ideally, this experimental setup would be altered to include the necessary brake and gas pedals. In terms of a brake reaction time, the hand response shows no kinematic difference when compared to foot pedal responses. There may be differences in reaction times between these two, but those differences can't truly be measured until we have data regarding the movement of the lower legs, and in this instance, the upper extremities as well.

With all this being said, no driving simulator will be able to give the same kind of feedback as an on-road test. Improvements can, however, be made in order to more accurately mimic the kind of situations a driver may experience on the open road.

Distractions come from everywhere while driving- in the rear view mirror, the periphery of the driver's view, through the driver's ears, and even in the back seat. The best clinicians and therapists can do at this point is utilize the technologies available to them to assess the ability of an individual to drive.

If a method could be developed and implemented to more closely assess the kinematic movement of the driver's lower limbs, researchers and clinical therapists could be provided with more information on how to best rehabilitate drivers. All of the above brake reaction timers used in previous work did an excellent job of assessing brake reaction time as a stand-alone variable. However, this picture of driving is only half complete. If kinematic data can be provided, the not only will the "what" of driving be answered, but more importantly, the "how" can be quantitatively measured.

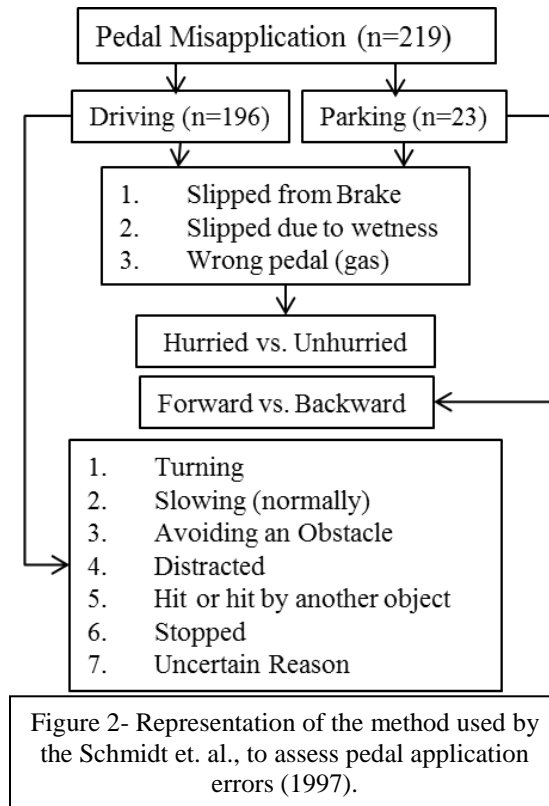
Pedal Application Errors

While driving a car is more than just seeing brake lights or responding to emergency situations, these are the most crucial situations for any driver. But what happens when the responses to these stimuli are inadequate or inappropriate? Many of the car accidents happen as a result of pedal application error- not simply distracted driving. Errors resulting in damage or injury have been documented by police reports across the country, and researchers have used this data to try and figure out how these errors occur and why¹⁸. In a number of studies conducted since the 1980s, several different types of pedal application errors have been analyzed and researched. The first of these defined pedal errors was termed "unintended acceleration" in the 1980s, when a rise in the number of crashes at the start of a driving cycle increased to the point that a "60 Minutes"

feature was done on it. Mechanical and automotive assessments were often performed on these vehicles which had a crash, and no problems were ever found that would be associated with this unintended acceleration¹⁹. The most likely cause of these accelerations is pedal misapplication, when the driver depresses the gas pedal when they intend to use the brake. They then press down harder trying to stop the car, which causes a rapid and violent acceleration with the driver unable to stop the car before causing damage or injury to the driver or others¹⁸. The attempted depression of the brake was done to prevent “creeping” of the vehicle, but applying the gas pedal instead causes a motor vehicle accident. It was hypothesized that the driver became misaligned in one way or another at the onset of driving, and this “pedal misapplication” all but disappears in typical driving, reemerging at venues such as drive up ATMs or fast food drive thrus¹⁸. Upon the conduction of this study, however, these misapplications were found to be more prevalent than originally thought. Of over one million reported accidents of varying degrees from North Carolina Police Accident Reports made from 1979-1980, 219 were found to be the direct result of obvious pedal misapplications of one type or another as recorded by the police reports, the verification of which can’t be done without visual or kinematic evidence. None of these incidents occurred at the start of the driving cycle (initial shift from park to drive). The task then came to breaking down each of these driving errors¹⁸.

Driving errors as found by Schmidt et. al. were classified in the manner indicated in Figure 2. As shown in the figure, only 23 of the 219 total incidents were found to be as a result of errors in “parking”, which was defined as driving in areas such as parking lots,

garages, or street side parking. Driving was defined as all other environments, residential, urban, rural, and highway. A distinction was made between foot slipping due to a wet foot pedal or shoe, as compared to a slip from other reasons.



The “uncertain reason” selection (#7 of driving errors), was selected in instances when the report was not specific enough to allow the researchers to come to a reasonable conclusion as to why the error was made¹⁸.

The findings from the parking errors section shows that majority of the crashes were the result of “wrong pedal” depression (n=15), Forward direction (n=14), and in unhurried circumstances (n=18). Driving situation conclusions vary from those of parking situations, with 110 of the 196 incidents coming as a result of foot slipping, 117 in unhurried situations, and 56 (more than in any other) in slowing situations, as in

approaching traffic¹⁸. A second Schmidt study performed in 2010 with a larger sample size confirmed these results, stating that these pedal misapplications occur not only at the start of the driving cycle, but also throughout various driving situations¹⁹.

The specific variable sought out by the authors of this paper is most like the movement time variable as defined by Warshawsky-Livne et. al. (2002). To give a basal level of ability, healthy drivers between the ages of 18 and 26 were selected to perform a battery of tests to try to quantify movement time and possible correlations with pedal application errors⁷. To induce these mistakes and show differences between typical and atypical driving, the test subjects were asked to perform trials with the right foot and the left foot. Driving with the left foot is something that is not regularly practiced, and even the use of the clutch pedal in manual transmission vehicles doesn't mimic the kinds of motions required to move back and forth from the gas and brake pedal.

Lower Extremity Kinematic Chain

To understand how the various methods of driver testing work, the kinematic chain of the lower limbs must first be defined. The term "kinematic chain" refers to the joints involved in pedal operation, as well as the muscles that control them. While the study conducted did not explore the muscle activation patterns specifically, certain speculations and conclusions can be made based on existing work and literature.

Driving a car smoothly takes a tremendous amount of finite muscle control- which is why driving with the left foot could be perceived as being more difficult for most people. The several muscles involved in gas and brake pedal application have to work in concert and in an appropriate proportion. Otherwise the ride is not smooth, and

applying the pedals may prove to be more difficult than it should be. The muscle activation is seemingly simultaneous with all of the joints when foot movement is required. However, further characterization needs to be done to thoroughly assess which joints are moving when, and in what proportion. Analyzing the kinematic chain in healthy, competent drivers can help to characterize what atypical or otherwise unsafe movements may look like. The joints in question include the ankle/foot complex (AFC), knee, and hip.

The AFC joint is fairly complex, and doesn't have a conventional bone articulation like many other joints, such as the carpals, knee and elbow. Rather, it is made up of several bones articulating with one another generating two major kinds of movement: inversion/eversion and plantar flexion/dorsiflexion²⁰. The Talus and Calcaneus bones of the foot, and the Tibia and Fibula of the lower leg all interact with one another to create these different motions. AFC eversion is defined as the AFC rotating externally, or away from the centerline of the body²⁰. This is performed by the three Peroneal muscles (longus, brevis, and tertius) working together to pull up the lateral aspect of the foot²¹. AFC inversion is, by contrast, an internal rotation, rolling the foot towards the centerline. Inversion is accomplished by the activation of the tibialis posterior, but also with the assistance of Flexor Digitorum Longus and Flexor Hallucis Longus²¹. In our context, when a driver is operating the pedals with the right foot, AFC inversion is likely to occur when the foot moves from the gas to the brake pedal, and eversion when the foot is moving from the brake back to the gas. The inversion and eversion of the AFC joint is most likely flipped when the driver is using the left foot.

The second set of motions found at the AFC joint complex is plantar flexion and dorsiflexion. Plantar flexion is when the end of the foot is most distal from the center of the body or pointed downward, and the angle of the AFC is increasing. Dorsiflexion is just the opposite, when the foot is pointed up and the angle of the AFC is decreasing²⁰. Plantarflexion is done by activating the gastrocnemius, soleus, and plantaris muscles²¹. Dorsiflexion is performed by the muscles found on the anterior of the lower leg, such as extensor digitorum longus, extensor hallucis longus, fibularis tertius, and the primary mover, tibialis anterior²¹. Plantar flexion is performed while driving when acceleration or braking is necessary, allowing the driver to depress the pedal to a greater degree. Dorsiflexion occurs when the driver picks their foot off of the pedal, or at least reduces the level of pedal depression.

While also capable of internal and external rotation, the knee's primary function is to flex and extend the lower leg. Flexion is defined as the motion in which the angle of knee is decreasing, or the lower leg is folding behind the upper leg, and extension is when the angle of the knee is increasing²⁰. Knee flexion is performed by and large by the muscle group in the back of the thigh, known as the hamstrings, though is assisted by the gastrocnemius (a biarticular muscle). The hamstrings are composed of three major muscles, the biceps femoris, the semimembranosus and the semitendinosus²¹. In the context of driving, these three muscles work together to bring the lower leg away from the gas and brake pedals, so a pedal shift may be made. Knee extension is performed by the muscle group known as the quadriceps, which includes the rectus femoris and the 3 vastus muscles, lateralis, medialis, and intermedius²¹.

The final joint in consideration while operating the gas and brake pedals of a car is the hip joint. While the hip is a ball and socket joint capable of several different kinds of motions including abduction and adduction and internal/external rotation²⁰, this study focused on and gathered information specifically regarding flexion and extension at the hip. Hip flexion is performed by muscles on the anterior aspect of the hip, which includes the adductor complex (brevis, longus, and magnus), the pectineus, and gracilis, as well as the rectus femoris once again²¹. Hip extension is performed primarily by the gluteus maximus, though other muscles also assist in the motion, such as the other gluteal muscles, medius and minimus²¹.

Because of the triaxial nature of the hip joint²⁰ (capable of moving in the three different planes of anatomical motion- transverse, sagittal, and frontal), the list of active muscles in any flexion or extension is quite extensive. For this reason only the primary movers will be listed. Very few motions at the hip are solely using the muscles required for just one plane of motion- the range of motion and joint mobility make the hip an extremely versatile joint, allowing us to perform many types of locomotion and different seated positions.

Neural pathways are another important variable to take note of. The way we control our limbs is developed through experience, and is perfected through a means known as the Neuronal Group Selection Theory, where cortical and subcortical systems of the brain are dynamically organized into variable networks, the organization of which is determined by development, behavior, and environment²². These networks accomplish very specific tasks in a specific way, maturing through aging.²² ²³. Expanding on this

idea of a cortical or neuromuscular effect, there are several diseases, disorders, or other ailments which may prove to change the drivers movements in a profound way. These diseases may include Parkinson's disease, Diabetes Mellitus, Stroke, limb amputation, implantation of an orthopaedic device, and several other degenerative diseases associated with the aging process. The roles of the central and peripheral nervous systems are yet another variable that may be assessed in future work by various means, including electromyography (EMG) instrumentation among other methods.

Current Study Objectives

Limited quantifiable data exists on lower extremity mobility and function during driving. Measuring the kinematic baseline values during driving may prove to be an invaluable tool in lower extremity function assessment in drivers. In this thesis our objectives are to 1) Develop a system of angle measuring goniometers use in a driving simulator whereby the motions across the hip, knee and AFC joints could be measured during active driving simulator scenarios, and 2) To validate this methodology with a study of lower extremity functional performance during left leg and right leg driving a brake reaction time driving scenario with active peripheral vision queuing.

MATERIALS AND METHODS

Study environment:

This study was conducted in the DriveSafety CDS-250 driving simulator located in Room 314 Brackett Hall on the Clemson University campus in Clemson, SC. The studies were performed during weekdays at daytime hours with the earliest test subject performing at 8am and the latest test subject completing by 9pm. All data collection was completed in a two week time span in May 2011. Drivers were asked to wear comfortable, closed-toe shoes and to provide their own spandex or tight-fitting compression shorts to allow for accurate placement of goniometers, and to ensure that shifting and noise in the data stream would be limited. If the subject did not have an appropriate pair of compression shorts or pants, a pair was provided for them. The driving simulator room was kept at room temperature, and other fans were placed around the room to maintain an environment that would be as comfortable as possible for the driver.

Subject Sampling:

To ensure that test subjects were of similar stature, the recruitment flyer detailed the specific sizes allowable for the study. In addition to being a male having a valid U.S. driver's license and at least one year active driving experience (defined as driving a car 5 times or more per week), the driver must also meet the sizing guidelines found in tables 2 and 3.

Units for recruitment were defined in standard units as opposed to metrics units due to familiarity of the potential subject pool with that measurement. For all statistical and mathematical calculations, units were converted back to metric.

BMI CALCULATOR CHART		
Height	Min. Wt.	Max Wt.
5'8"	124	164
5'9"	128	169
5'10"	132	174
5'11"	136	179
6'0"	140	184
6'1"	144	189
6'2"	148	194

Table 2: BMI calculation chart as shown on recruitment flyer.

Test Subject Requirements		
Condition	Min	Max
Height (ft.'in.)	5'8"	6'2"
BMI (Body Mass Index)	18.5	25
Shoe Size (US Sizes)	8	12
Age (Year)	18	29

Table 3: Conditions met by each test subject.

The BMI (standards for test subject inclusion found in table 2, as shown on the recruitment flyer) value minimum and maximum are based on the National Institute of Health measure for normal weight. All test subjects met the criteria, and the statistical information of the subjects (n=13) is found in Table 3. In addition to these inclusion criteria, the subjects were asked if they were capable of operating a manual transmission vehicle. We felt this was a relevant piece of information because it may show a predisposition to being able to drive with the left foot in a manner more closely mimicking the right foot before the start of the trials. Seven subjects responded in the affirmative, and six stated that they could not operate a manual transmission vehicle. Body segments were also measured using a tape measure in centimeters. A tape measure was used in order to assure that bends and contours due to muscles or bone prominences were limited, and the most absolute straight line was measured. This included the foot, lower leg, upper leg, torso, upper arm, and lower arm. Bony landmarks were located on

each of the test subjects through light palpation. The length of each segment is defined as follows:

1. *Foot: Heel to tip of shoe (shoe was on to measure length of foot in driving conditions)*
2. *Lower Leg: lateral malleolus to lateral condyle of the tibia*
3. *Upper Leg: lateral condyle of tibia to greater trochanter of the femur*
4. *Torso: greater trochanter of tibia to greater tuberosity of the humerus*
5. *Upper Arm: greater tuberosity of the humerus to olecranon process of the ulna*
6. *Lower Arm: olecranon process to styloid process of the ulna.*

The final length measured was the horizontal distance from the point of the hip of the seated driver to the front edge of the gas pedal. This was done by placing the tape measure next to the seat, and starting at a point directly adjacent to the gas pedal, the tape measure was extended to the point directly below the point of the hip on the floor of the driving simulator. This was done to give a preliminary understanding of how flexed or extended each segment would be relative to the absolute distance to the gas pedal.

Test Subject Instrumentation:

Each test subject was instrumented with six goniometers, designed and developed by Biometrics, Ltd., of Cwmfelinfach, Gwent, Wales. These goniometers were placed at the left and right AFCs, knees, and hips while the subject was standing. The goniometers on the AFCs were taped to the side of the shoe with the tip of the sensor even with the tubercle of the 5th metatarsal for most subjects, and the other end was attached to a Velcro

timing strap (used in running and multisport events) via Velcro tape. The goniometers at the knee were attached by connecting two timing straps together and connecting the goniometer in the same fashion as the AFC goniometer. For the hip sensors, the top of the sensor was taped to the spandex at the beltline with the tip extending to approximately the iliac crest, and the lower portion of the goniometer was taped along the seam of the shorts extending down the thigh. All of the sensors were attached on the lateral aspect of the legs, so as to obtain the most accurate movement in the sagittal planes as possible.



Figure 3: Biometrics placement using a different attachment technique- double-sided medical grade tape.

Once the goniometers were in place, the test subject was asked to sit in the driver's seat and any necessary adjustments to goniometer placement were made.

At 90-degree flexion for any of the joints, there should be no excess spring flexion or bulging. If such errors were made, the goniometers were moved to get rid of

the excess spring bends. After all the goniometers were in their appropriate place, the sensors were plugged in to the Biometrics computer and the sensors were zeroed when the subject was standing at anatomical position (feet shoulder width apart, arms at side, head up, standing up straight).

Driver Training:

Once the driver was completely instrumented and seated in the driving simulator, a basic description of the simulator was given. This included a description of the controls, gear shift, seat adjustment and seatbelt. After the driver became used to the environment, they then went through a battery of warm-up conditions, with no data being gathered. The different driving scenarios included lane awareness, speed maintenance, and functional object detection. Lane awareness was established using a 5-circle system, with the center circle lit up green when in the center of the lane, yellow lights to either side if the car was not in the center and red lights if the car was leaving the lane. All 5 lights would turn red if the vehicle completely left the lane. A car was then placed in the lane directly in front to allow the driver to get used to following a car, which would become important once data was being collected. The functional object detection task required the driver to stay in their lane (with the cruise control on) and use their peripheral vision to see “E” (figure 5(a)) markers to the left or right of center and press the corresponding button, found on the steering wheel (figure 5(b)). The presence of the “E” markers at the periphery of the test subject’s vision functions as a method to simulate typical scanning tasks which occur with on-road driving. Simply offering a brake light in front of the test subject causes an

oversimplification of the reaction time and movement time required for braking, so the action in the simulated environment aids in the recreation of a real driving scenario.



Figure 4: View of the CDS-250 Driving Simulator on Clemson University Campus.



Figure 5(a) (left): Virtual environment with target “E” on the screen. (b) (right) Steering wheel and red functional object detection buttons.

With the driver comfortable with the operation of the driving simulator, the test data collection would now begin. The tested scenario was a compilation of the training scenarios, requiring the driver to do many things at once. This includes maintaining a

speed of 55 mph +/- 10mph with the pedals, seeing and responding to “E” signals to the left and right of center in a timely manner, and responding to braking events signaled by brake lights of the vehicle ahead. Test subjects were asked to drive with both feet, one at a time, for 3 trials (total of 6). Test subjects with odd numbers (1, 3, 5, 7, 9, 11, and 13) were asked to perform the driving scenario with their right foot, while subjects with even numbers were asked to drive with their left foot first. After trial 2, the drivers were asked to stand up to allow for re-zeroing of the sensors, or adjust the goniometers should they have moved during data collection.

These goniometers placed at the AFC, knee, and hip of both legs gathered data reflecting the joint angle relative to the values found at anatomical position. The goniometers are precise to the tenth of a degree, and are reported to have an accuracy of ± 2 degrees for measures under 90 degrees, and ± 3 degrees for measures over 90 degrees. Data for the joints was collected at a sampling rate of 200Hz. All goniometric data was collected using the default orientations and sign conventions of the sensors when placed on the individual joints. With respect to the right leg, the sensors recorded increasing positive angles with AFC Inversion, AFC Plantar-Flexion, Knee Flexion, and Hip Flexion. For the left leg, the sensors recorded increasing positive angles with AFC Inversion, AFC Dorsiflexion, Knee Extension, and Hip Extension. The reversal of raw sign convention between the left and right legs is a product of the reversal of the orientation of the goniometer systems with respect to bending. During post processing, all measures were converted to result in a universal sign convention, whereby positive

increasing angles were seen for AFC Inversion, AFC Plantar-Extension, Knee Flexion, and Hip Flexion.

In addition to the goniometric data, the values of the brake response time and the time for the brake to gas movement. The brake response time is defined as the time elapsed from the brake lights of the virtual car lighting up to a 5% depression of the brake pedal. The 5% depression value is measured via potentiometer, and that value triggers the brake lights of the virtual car to turn off. The movement time from brake to gas was defined as the time elapsed from brake pedal release to 5% gas pedal application after the braking response. Drive Simulator data was collected at a 60Hz sampling rate, so in instances when 5% application was not met at a sample, data interpolation was used to find where the mark would have been hit.

During the course of testing, the driver was asked a series of questions to assess simulator sickness symptoms they were experiencing. This ranking system assured the tester that the test subject was not undergoing any unnecessary stress. This questionnaire was a series of 17 conditions in which the subject answered how they were feeling in regards to that condition on a scale of 0 to 10, with 0 being not at all and 10 being very. Some of these symptoms included queasy, uneasy, hot, sweaty, nauseous, floating, and many other similar conditions that may arise from being in the driving simulator and reflect motion sickness.

Laboratory Process and Statistical Analysis:

The following secondary measures were calculated during this study:

1. Brake response time- time elapsed from brake stimulus (brake lights of virtual car lighting up) to 5% brake application
2. Brake to gas movement time- time elapsed from brake pedal release after a braking response to 5% application of the gas pedal.
3. Minimum value of each of the following joints in the given motion:
 - a. Flexion and extension at the hip.
 - b. Flexion and extension at the knee.
 - c. Inversion/eversion at the AFC.
 - d. Plantar flexion/dorsiflexion at the AFC.
4. Maximum value of each joint in 3a through 3d.
5. Range of motion of each joint in 3a through 3d.
6. Slope values in terms of degrees of motion at joint “x” to degrees of motion at joint “y”. These values were be compared as (x axis joint listed first):
 - a. AFC Inversion/Eversion vs. AFC Plantar Flexion/Dorsiflexion
 - b. AFC Inversion/Eversion vs. Knee Flexion/Extension
 - c. AFC Inversion/Eversion vs. Hip Flexion/Extension
 - d. AFC Plantar Flexion/Dorsiflexion vs. Knee Flexion/Extension
 - e. AFC Plantar Flexion/Dorsiflexion vs. Hip Flexion/Extension
 - f. Knee Flexion/Extension vs. Hip Flexion/Extension
7. Correlation values for the given slopes for each of the joint comparisons in 1a through 1f.

The Biometrics data was processed in a way that allowed for the assessment of joint correlation, comparing each of the joint motions captured to each other. The joint correlations and slopes, as well as the individual values of the minimum, maximum and range of each joint was performed in MATLAB, using a novel program developed by author J. Tarbuton (program code located in Appendix A). Each of the 4 joint channels

(two axes at the AFC, one axis at both the knee and hip) were correlated to each other, yielding a total of 6 different sets of values for correlations and slopes. The variables were collected from brake stimulus to brake reaction, as well as brake release to gas application. In this manner, the movements to and from the gas pedal were characterized. This is because with the format of the study, the foot position immediately prior to brake stimulus cannot be ensured, thus causing a significant amount of variability in an uncontrolled part of the study. The correlations at brake application and brake release were collected as follows, in each of the six trials (three with each foot):

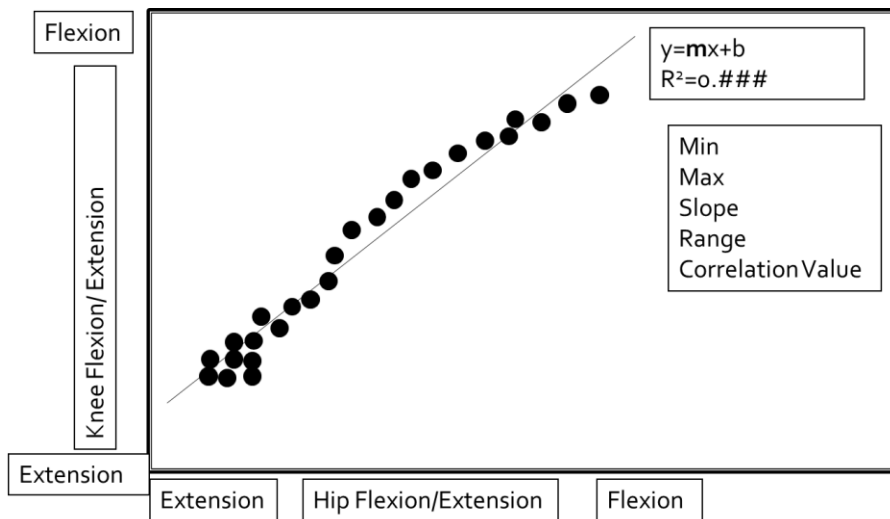


Figure 6: Graphical interpretation of what the data is proposed to look like upon processing

The data can be described graphically in the schematic shown here (Figure 6). From this figure, the values will be extracted for each braking response, and averages across the sample population for both the brake application and the return to the gas pedal. Given this information, t-tests will be performed to find where significant changes are made, assessing the left vs. right leg driving at the beginning and end of the trials, as well as the changes made in the left leg and right leg from beginning to the end of the session

(comparing 1st right vs. 1st left, 1st right vs. 3rd right, and 1st left vs. 3rd left, and 3rd right vs. 3rd left). The driving simulator data will be analyzed for statistical significance, comparing brake reaction times and brake to gas movement times in the same manner as the other parameters.

Brake responses that elicit an incorrect response or a pedal misapplication were considered in another aspect of analysis. The braking errors were quantified, and comparisons were made in the same fashion as the correlations, slopes, etc. The errors were defined in this study in the following manner:

1. Increased gas application- after the brake stimulus, the gas pedal is depressed to a level greater than that of the level at the stimulus. These errors are only counted if the increased application is equal to or greater than 5% when compared to the level of depression at the brake stimulus.
2. Gas Peak After Stimulus- in response to the brake stimulus, the gas pedal is mistakenly applied, followed by the correct braking response. This error will manifest as an independent gas peak, starting after the braking stimulus.
3. Double depression- in reaction to a brake stimulus, both the brake and gas pedal are applied simultaneously in error.

With these 3 basic pedal application errors, most of the possible erroneous events are described and may be quantified in a manner that can aid in the assessment of safe driving, and show how errors occur and what the effective result of those errors may be.

With the assistance of J. Sharp of the Clemson University Statistics Department, a custom SAS (version 9.2) statistical script (function PROC TTEST) was developed to

conduct a multiple ANOVA statistical comparison for all variables of interest between trials and leg. Post-hoc t-tests were performed to assign statistical significance at a confidence of 0.05. These t-tests were based off a matched pairs design, utilizing the comparison of left vs. right, and first vs. third trials.

RESULTS

Subject Demographic Data

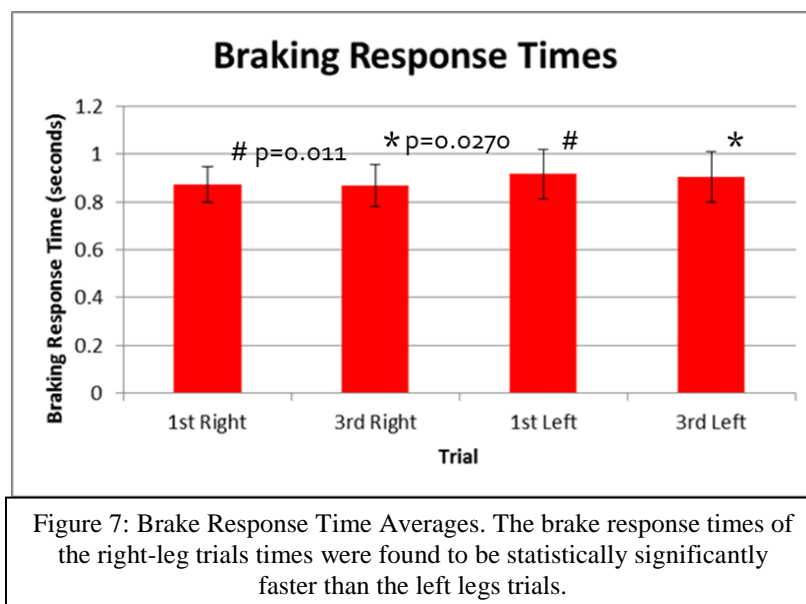
The table below (figure 4) shows the test subject anthropometric data. This shows that the test subjects were within narrow confines of body dimensions, and very little variability was found with regard to body posture. This data was not used any further in the current study.

	Average	Stand. Dev	Max	Min	Range
Ht (cm)	180	5.81	187.96	172.72	15.24
Weight (kg)	71.9	5.04	81.63	62.583	19.047
BMI (kg/m ²)	22.21	1.5	24.95	20.05	42.533
Shoe Size	10.77	1.15	12	8.5	3.5
Age	21	2.2	26	19	7
Foot Length (cm)	24.85	0.9	26	23	3
Lower Leg (cm)	43.85	2.03	48	41	7
Upper Leg (cm)	44.54	3.43	50	39	11
Torso (cm)	56.69	3.45	62	50	12
Upper Arm (cm)	35.77	2.39	40	33	7
Lower Arm (cm)	28.12	1.33	29	25	4
Pedal-->Hip (cm)	91.69	2.93	98	88	10

Table 4: Data collected from the body dimensions of the test subjects.

Brake Reaction Time Data

Average and standard deviation values for each trial and leg for the measure of brake reaction time are displayed in figure 7 below, and also found in tabular form in table 4 below. For Brake Reaction Times (BRT), the right BRT was found to be statistically significantly (SS) faster than the left BRT in trial 1, with right BRT showing .87+/-0.08 seconds and left BRT showing .92+/-0.10 (p=0.0110).



This statistical difference remained over the testing, with trial 3 reaction times showing that the right BRT (.87+/-0.09) was SS faster than the left BRT (.91+/-0.10)(p=0.0274). Neither the left nor the right leg showed SS changes in these reaction times between trial 1 and 3, with BRTs remaining similar between trial 1 and 3 for the Left Leg (.92+/-0.10 and .91+/-0.10, respectively) and the right leg (.87+/-0.08 and .87+/-0.09, respectively).

Brake Reaction Time Lower Extremity Kinematic Data

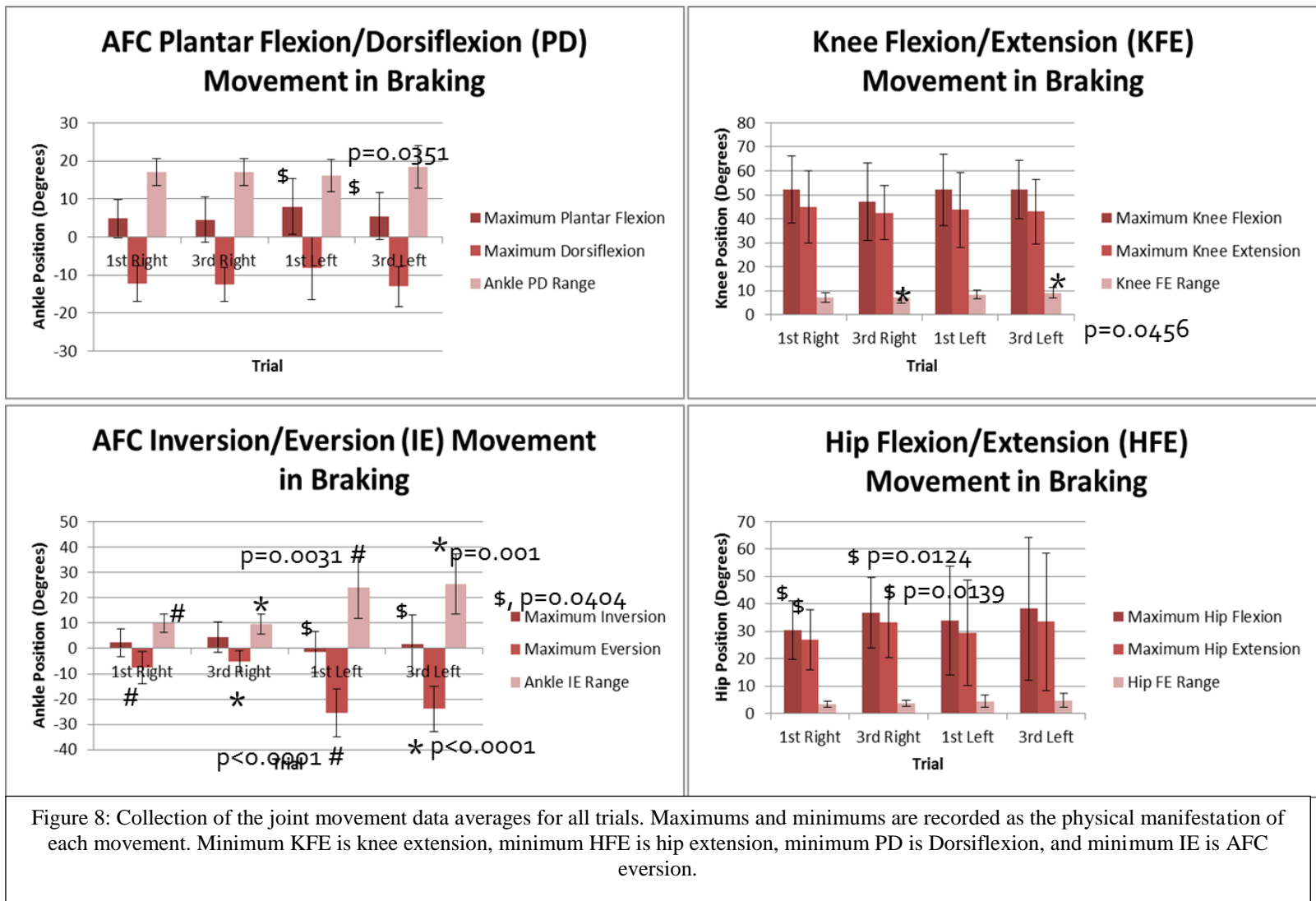
The table (table 5) and figure (figure 8) below show the average and standard deviation values for each trial of each measured variable of the study. A symbol notates the statistical significance with a '\$' indicating that there is a statistically significant difference (SSD) between trial 1 and trial 3 for either the right or left leg. The symbol '#' represents a difference when comparing the first trials of the left and right. The final symbol used was the '*', which indicated that there was a SSD found when comparing trial 3 of the left and trial 3 of the right leg.

DATA AND STATISTICAL ANALYSIS FOR BRAKE RESPONSE PHASE OF AUTOMOBILE OPERATION								
Parameter	Trial	Maximum (°)	Minimum (°)	Range (°)	Parameter	Trial	Correlation	Slope (joint y°/joint x°)
AFC Plantar Flexion & Dorsiflexion	1st Right	4.8±5.1	-12.2±4.6	17.1±3.5	KFE/PD	1st Right	0.7±0.2	-2.6±0.8 #
	3rd Right	4.5±6.0	-12.5±4.5	17.0±3.6		3rd Right	0.7±.2	-2.4±1.2
	1st Left	8.0±7.3 §	-8.2±8.3	16.2±4.2		1st Left	0.7±0.2	-1.8±0.9 #
	3rd Left	5.4±6.2 §	-13.0±5.3	18.4±5.5		3rd Left	0.8±0.2	-2.0±1.0
AFC Inversion & Eversion	1st Right	2.3±5.6	-7.6±6.4 #	9.9±3.6 #	HFE/PD	1st Right	0.5±0.3	-3.5±2.7
	3rd Right	4.5±6.0	-5.1±4.2 *	9.5±4.0 *		3rd Right	0.5±0.6 *	-3.8±3.0
	1st Left	-1.5±8.1 §	-25.4±9.5 #	23.9±12.0 #		1st Left	0.6±0.2 §	-3.8±2.8
	3rd Left	1.6±11.6 §	-23.8±8.9 *	25.4±11.9 *		3rd Left	0.8±0.1 *§	-4.7±4.1
Knee Flexion & Extension	1st Right	52.1±14.0 §	45.0±15.0	7.2±1.9	KFE/IE	1st Right	0.3±0.2 #	-0.1±0.7 #
	3rd Right	47.2±16.1 §	42.5±11.3	7.3±2.4 *		3rd Right	0.3±0.2 *	-0.3±0.7 *
	1st Left	52.0±15.0	43.7±15.6	8.3±1.9		1st Left	0.8±0.2 #	-2.7±1.6 #
	3rd Left	52.2±12.1	43.1±13.4	9.1±2.3 *		3rd Left	0.8±0.2 *	-2.8±1.8 *
Hip Flexion & Extension	1st Right	30.4±10.7 §	26.9±11.0 §	3.5±1.1	HFE/IE	1st Right	0.4±0.3	-0.8±2.1 #
	3rd Right	36.9±12.9 §	33.3±12.9 §	3.6±1.2		3rd Right	0.3±0.2 *	-0.5±1.6 *
	1st Left	33.9±19.8	29.4±19.2	4.5±2.1		1st Left	0.6±0.3	-5.3±4.4 #
	3rd Left	38.3±26.0	33.5±25.1	4.8±2.5		3rd Left	0.7±0.2 *	-5.8±5.6 *
Brake Response Time		Time (s)			HFE/KFE	1st Right	0.5±0.3	1.7±1.1
	1st Right	0.87±0.08 #				3rd Right	0.5±0.3 *	1.6±1.3
	3rd Right	0.87±0.09 *				1st Left	0.7±0.3	1.8±0.9
	1st Left	0.92±0.10 #				3rd Left	0.8±0.1 *	2.0±1.7
Statistically Significant Difference (SSD) at α=.05					PD/IE	1st Right	0.3±0.2 #	-0.1±0.2 #
						3rd Right	0.3±0.6 *	-0.0±0.2 *
						1st Left	0.6±0.3 #§	1.0±0.6 #
						3rd Left	0.6±0.3 *§	1.1±0.5 *
		*- SSD 3vs3	§- SSD 1vs3	#- SSD 1vs1				

Table 5: Brake Response data table. Knee Flexion/Extension (KFE), AFC Plantar Flexion/Dorsiflexion (PD), AFC Inversion/Eversion (IE), and Hip Flexion/Extension (HFE)

During braking tasks, the lower extremity motions (min, max and range) for each joint were measured. The maximum value of plantar flexion/dorsiflexion at the AFC, or the greatest average degree of dorsiflexion (toes towards the body) was found to be SSD

between the first vs. third trial on the left leg (8.0 ± 7.3 and 5.4 ± 6.2 , respectively ($p=0.0351$)). This same trend is found in the maximum value of eversion- outward rotation of the AFC- in the first and third trials of left leg (-1.5 ± 8.1 and 1.6 ± 11.6 , respectively ($p=0.0404$)).



With respect to the right leg, SSDs were found in the first and third trials of braking maximum knee flexion (52.1 ± 14.0 and 47.2 ± 16.1 , $p=0.0456$), maximum hip flexion (30.4 ± 10.7 and 36.9 ± 12.7 , $p=0.0124$), and hip flexion minimum, or where the hip was most extended (27.0 ± 11.0 and 33.3 ± 12.9 , respectively ($p=0.0139$)). No SSD existed in the fields of slope and correlation values when comparing right leg driving at the beginning and end of the session.

The slope values reported compared different joints to each other, and the numerical value of these slopes had the significance of relative joint motion as found in the table below (Table 6).

HFE/KFE=	<u>degrees motion at knee</u>	KFE/IE=	<u>degrees motion at AFC I/E</u>
	degrees motion at hip		degrees motion at knee
KFE/PD=	<u>degrees motion at AFC P/D</u>	HFE/IE=	<u>degrees motion AFC I/E</u>
	degrees motion at knee		degrees motion hip
HFE/PD=	<u>degrees motion at AFC P/D</u>	PD/IE=	<u>degrees motion at AFC I/E</u>
	degrees motion at hip		degrees motion at AFC P/D

Table 6: Listing of the slopes as they should be interpreted.

The correlation coefficients of these slopes indicate the deviation of the data within these slope curves. For trial 1, SSD differences in correlation between the left and right legs were noted between PD vs. IE and IE vs. KFE. For right leg driving, SSD differences between trial 1 and 3 were found between many more measures, including PD vs IE, PD vs. HFE, IE vs KFE, IE vs. HFE, and KFE vs HFE. Correlation values for HFE vs. PD (0.6 ± 0.2 and 0.8 ± 0.1 , first and third ($p=0.0064$)) and PD vs. IE (0.6 ± 0.3 and 0.6 ± 0.3 , first and third ($p=0.0426$)) also showed this SSD.

When comparing across the joints, looking at right vs. left, we see several differences at various parameters. One of the most pronounced differences was found in

the AFC inversion/eversion parameter both in the first and third trials. SSDs were found in the range (first trial, 9.9 ± 3.6 right and 23.9 ± 12.0 left, $p=0.0031$; third trial, 9.5 ± 4.0 right and 25.4 ± 11.9 left, $p=0.0010$) and minimum value (first trial, -7.6 ± 6.4 right and -25.4 ± 9.5 left, $p<.0001$; third trial, -5.1 ± 4.2 right and -23.8 ± 8.9 left, $p<.0001$), which indicates that the left leg performed much more AFC eversion to accomplish the braking task. This result was expected, as AFC eversion was predicted to occur to a larger degree while driving with the left leg as compared to the right leg.

Statistically significant differences were also found when comparing the right and left legs in several slope and correlation values. One of these SSDs was found in the first trial of KFE vs. PD slope (-2.6 ± 0.8 right and -1.8 ± 0.9 left, $p=0.0096$), while many others had SSDs in both the first and third trial comparisons of right and left. KFE/IE slope (first trial, -0.1 ± 0.7 right -2.7 ± 1.6 left, $p=0.0010$; third trial, -0.3 ± 0.7 right -2.8 ± 1.8 left, $p=0.0014$) and correlation (first trial, 0.3 ± 0.2 right and 0.8 ± 0.2 left, $p<.0001$; third trial, 0.3 ± 0.2 right and 0.8 ± 0.2 left, $p<.0001$), both had SSD when comparing right and left leg. This is to be expected, as the left AFC is performing different motions in the secondary axis- or AFC inversion/eversion- than the right leg. Other variables which include IE The variables found to have a SSD include HFE/IE slope (first trial, -0.8 ± 2.1 right and -5.3 ± 4.4 left, $p=0.0157$; third trial, -0.5 ± 1.6 right and -5.8 ± 5.6 left, $p=0.0114$), and PD/IE slope (first trial, -0.1 ± 0.2 right and 1.0 ± 0.6 left, $p<.0001$; third trial -0.0 ± 0.2 right and 1.1 ± 0.5 left, $p<.0001$) and correlation (first trial, 0.3 ± 0.2 right and 0.6 ± 0.3 left, $p=0.0007$; third trial, 0.3 ± 0.6 right and 0.6 ± 0.3 left, $p=0.0027$). Differences in third trial slope and correlation values were found between right and left in HFE/PD correlation

(0.5 ± 0.6 and 0.8 ± 0.1 , respectively, $p=0.0015$), HFE/IE correlation (0.3 ± 0.2 and 0.7 ± 0.2 , respectively, $p=0.0033$), and HFE/KFE correlation (0.5 ± 0.3 and 0.8 ± 0.1 , respectively, $p=0.0033$). The data show that most of the SSDs found with regard to the slope and correlation were found when comparing joints to the inversion/eversion movement at the AFC. This is due to the fact that as the leg goes to depress the brake pedal, eversion is required to center the left foot over the pedal, while inversion-or no significant motion at all- is required to center the right foot. The slopes across all the joints involving AFC inversion are less than -1, with some (such as plantar/dorsiflexion) approaching no relative motion.

No SSDs were found in the parameters of PD minimum and range, KFE minimum, HFE range, KFE/PD correlation value, HFE/PD slope, and HFE/KFE slope.

Based on the figures below (figures 14 and 15), some very obvious trends emerge. In both left-footed driving, the variables of KFE vs. PD, HFE vs. PD, and HFE vs. KFE all have very linear relationships, though the slopes may differ, indicating different positioning and movements in different proportions. When comparing PD vs. IE, KFE vs. IE, and HFE vs. IE, however, the left and right legs have drastically different movement patterns. None of these relationships appear to be linear, especially in the left leg. In KFE vs. IE and HFE vs. IE, the right leg shows some very subtle downward curving in its relationship. This variance from linear is much more prevalent when considering the left leg, with a dynamic curve representing the increased motion occurring at AFC inversion/eversion. It is clear through the observation of these figures that many motions

are occurring at once, and the smaller motions of the right leg are manifested in less dynamic curves with respect to inversion/eversion comparisons.

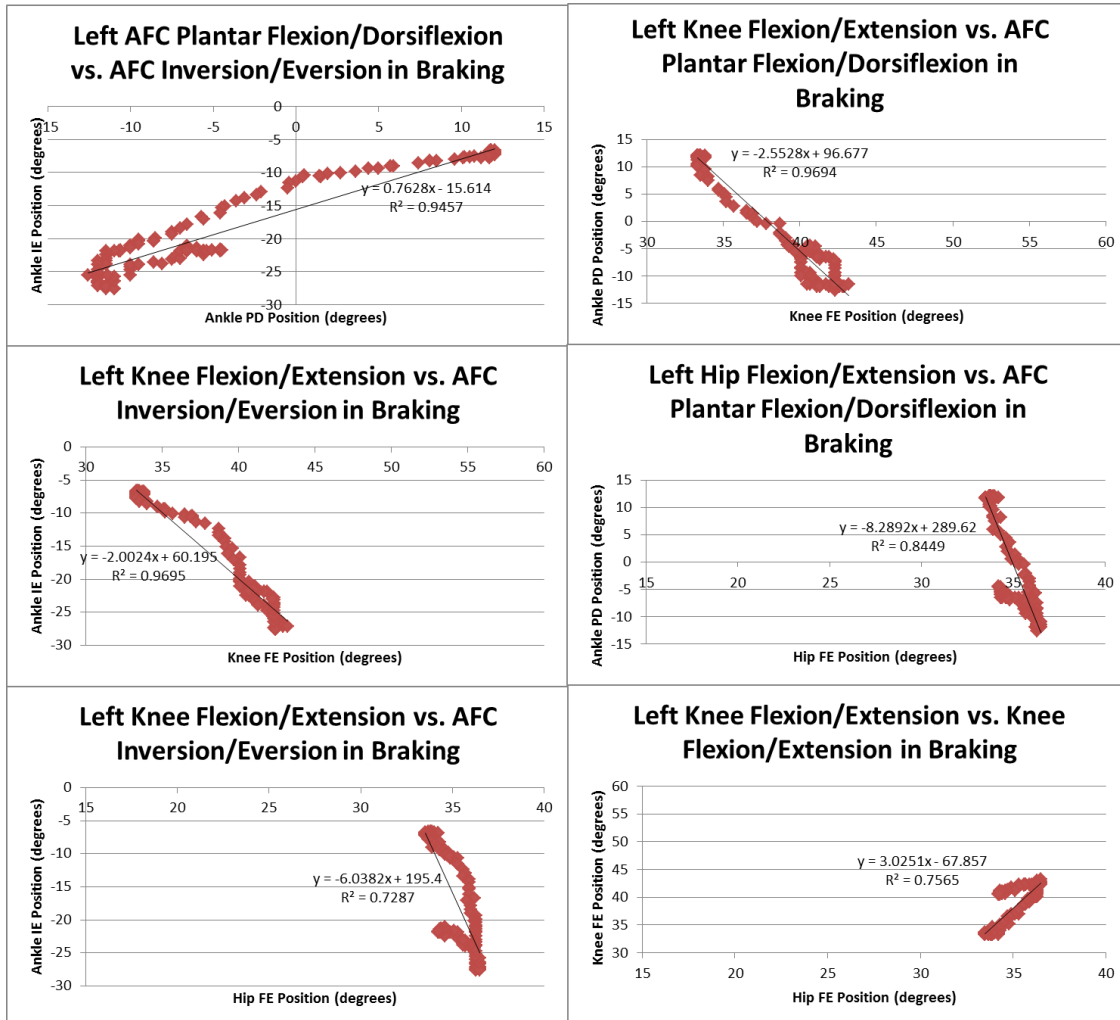


Figure 9: Collection of movement comparisons of left leg driving while braking. We see very linear relationships between joints not involving AFC inversion/eversion, while those involving the inversion/eversion have very distinct patterns to them, not appearing to have a linear relationship by any means.

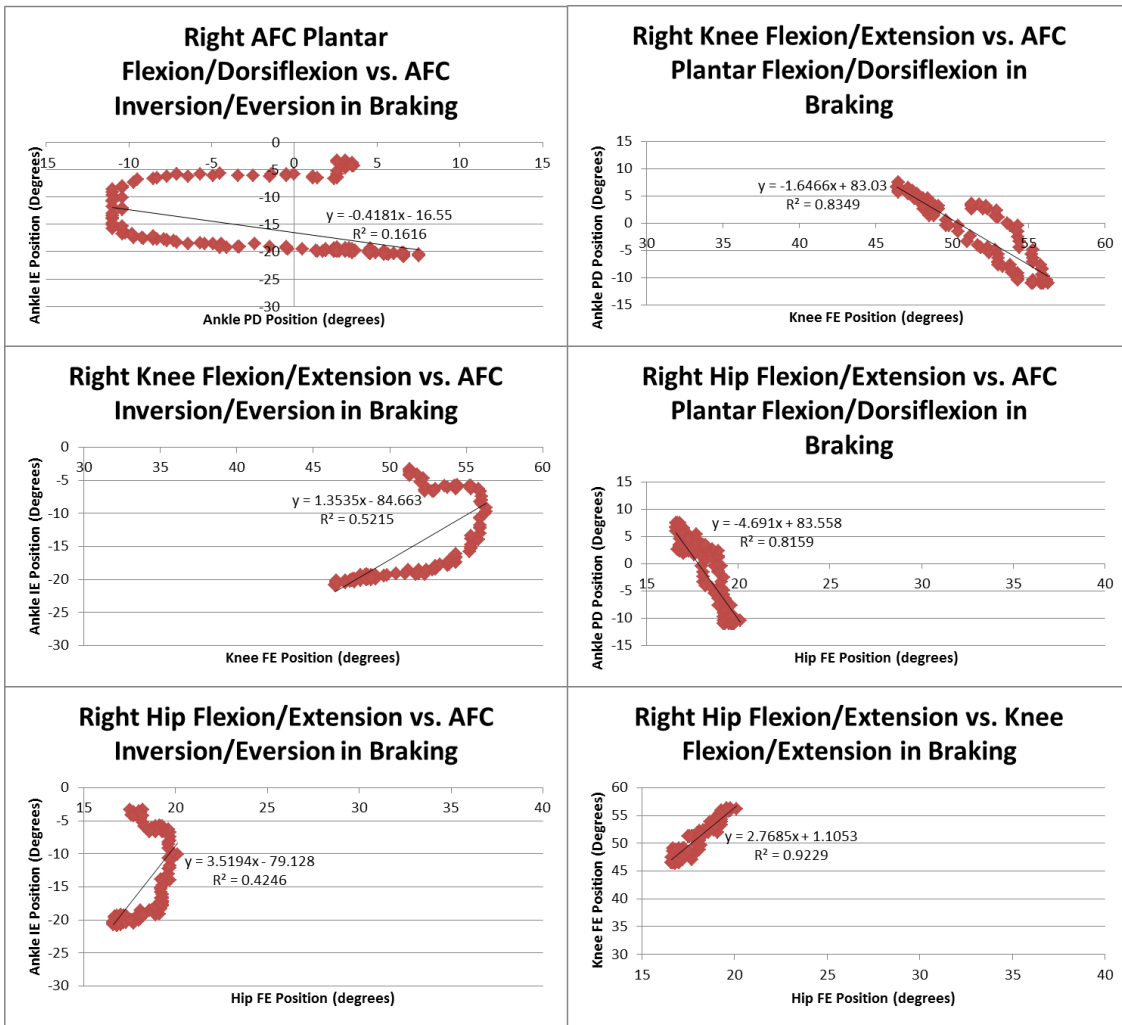


Figure 10: Right-leg braking response slopes and correlations. Note the linearity seen in comparisons not involving AFC inversion/eversion. Joints involving AFC inversion/eversion are less linear, especially AFC plantar flexion/dorsiflexion vs. inversion/eversion. The lack of linearity involved in inversion/eversion comparisons is the representation of multiple motions- two or more phases of motion- are incorporated with the movement from gas to brake pedal.

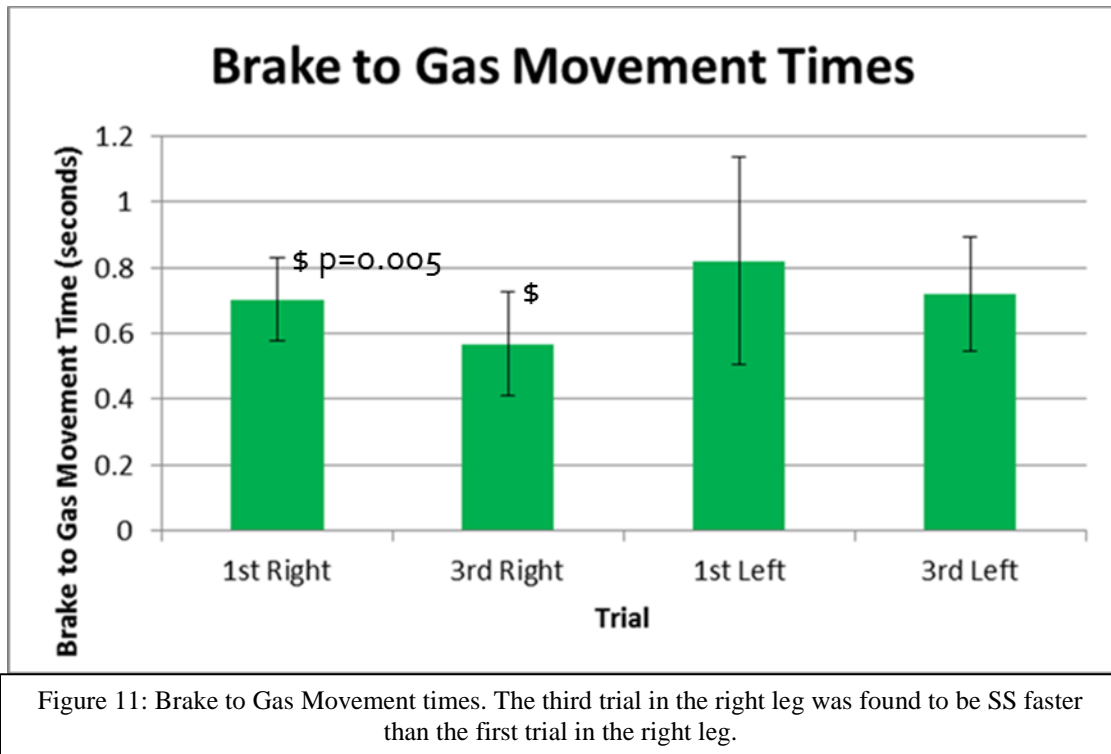
Return to Gas Times

On the opposite side of the braking response, the foot returns to the gas pedal with very similar instances of SSDs. Average and standard deviation values for each trial and leg for the measure of return to gas (RTG) are displayed in figure 12 below, and also found in tabular form in table 7 below. Movement time (figure 11) was found to be SS

faster in the third trial of the right leg when compared to the first trial (0.70+/-0.13 in the first trial, 0.57+/-0.16 in the third, p=0.0050). The comparison of other parameters of first and third trials on the same leg yields similar results to that of the brake application.

DATA AND STATISTICAL ANALYSIS FOR RETURN FROM BRAKE TO GAS PHASE OF AUTOMOBILE OPERATION								
Parameter	Trial	Maximum (°)	Minimum (°)	Range (°)	Parameter	Trial	Correlation	Slope (joint y°/joint x°)
AFC Plantar Flexion & Dorsiflexion	1st Right	3.6±4.5	-11.8±5.3	15.3±2.6	KFE/PD	1st Right	0.8±0.2	-3.1±1.5 #
	3rd Right	2.9±5.5	-11.9±5.2	14.8±1.9		3rd Right	0.8±0.1	-3.4±1.5 *
	1st Left	7.3±7.8	-6.6±8.6	13.9±3.4		1st Left	0.8±0.2	-2.1±1.2 #
	3rd Left	4.0±5.8	-10.9±5.7	14.8±4.7		3rd Left	0.9±0.1	-2.1±0.9 *
AFC Inversion & Eversion	1st Right	-1.7±5.3	-8.6±6.2 #	6.9±3.4 #	HFE/PD	1st Right	0.6±0.2	-3.0±4.3
	3rd Right	0.8±4.6	-5.8±4.1 *	6.6±3.3 *		3rd Right	0.5±.3 *	-2.4±5.4
	1st Left	-1.7±7.6 §	-20.9±7.4 #	18.9±10.4 #		1st Left	0.6±0.2 §	-3.3±2.2
	3rd Left	1.7±11.0 §	-18.1±11.1 *	19.8±10.0 *		3rd Left	0.8±0.1 *§	-4.2±3.1
Knee Flexion & Extension	1st Right	51.8±13.9 §	35.6±31.9	5.9±1.9	KFE/IE	1st Right	0.4±0.3	-0.5±0.7 #
	3rd Right	46.8±16.5 §	44.1±11.2	5.4±2.6 *		3rd Right	0.4±0.2 *	-0.5±0.5 *
	1st Left	51.2±15.1	43.9±15.6	7.1±2.0		1st Left	0.6±0.2	-2.0±1.7 #
	3rd Left	51.1±12.5	43.5±13.3	7.4±2.3 *		3rd Left	0.6±0.2 *	-2.3±1.5 *
Hip Flexion & Extension	1st Right	30.3±10.4 §	27.5±10.9 §	2.8±1.0	HFE/IE	1st Right	0.3±0.2 #	-0.5±0.8 #
	3rd Right	36.6±12.6 §	34.2±13.1 §	2.4±1.0		3rd Right	0.3±0.3 *	-0.9±1.0 *
	1st Left	33.7±20.6	29.9±20.0	3.8±1.8		1st Left	0.5±.2 #	-4.3±4.7 #
	3rd Left	37.9±26.4	34.0±25.4	3.8±2.2		3rd Left	0.6±0.2 *	-4.7±4.7 *
	Time				HFE/KFE	1st Right	0.5±0.3 #	1.2±1.1
Brake to Gas Movement Time	1st Right	0.70±0.13 §				3rd Right	0.5±0.3 *	1.3±1.7
	3rd Right	0.57±0.16 §				1st Left	0.7±0.2 #	1.5±1.1
	1st Left	0.82±0.32				3rd Left	0.8±0.1 *	1.9±1.4
	3rd Left	0.72±0.17			PD/IE	1st Right	0.5±0.3	0.2±0.2 #
Statistically Significant Difference (SSD) at α=.05						3rd Right	0.4±0.3	0.1±0.2 *
	§-SSD1vs3	#- SSD 1vs1	*- SSD 3vs3			1st Left	0.5±0.2	0.9±0.6 #
						3rd Left	0.6±0.2	1.0±0.5 *

Table 7: Return from Brake to Gas data table. Knee Flexion/Extension (KFE), AFC Plantar Flexion/Dorsiflexion (PD), AFC Inversion/Eversion (IE), and Hip Flexion/Extension (HFE)



Return to Gas Time Lower Extremity Kinematic Data

The figures below show the average and standard deviation values for each trial of each measured variable. In the right leg, we see SSDs from first to third trials in maximum knee flexion (51.8 ± 13.9 and 46.8 ± 16.5 , respectively ($p=0.0498$)), and hip flexion (30.3 ± 10.4 and 36.6 ± 12.6 , respectively, $p=0.0133$) and extension (27.5 ± 10.9 and 34.2 ± 13.1 , respectively ($p=0.0083$)). The left leg has SSDs from first to third trials in maximum inversion (-1.7 ± 7.6 and 1.7 ± 11.0 , respectively ($p=0.0311$)) and HFE/PD correlation (0.6 ± 0.2 and 0.8 ± 0.1 , respectively ($p=0.0055$)). No statistically significant differences were found comparing the first and third right-leg trial correlation or slope values.

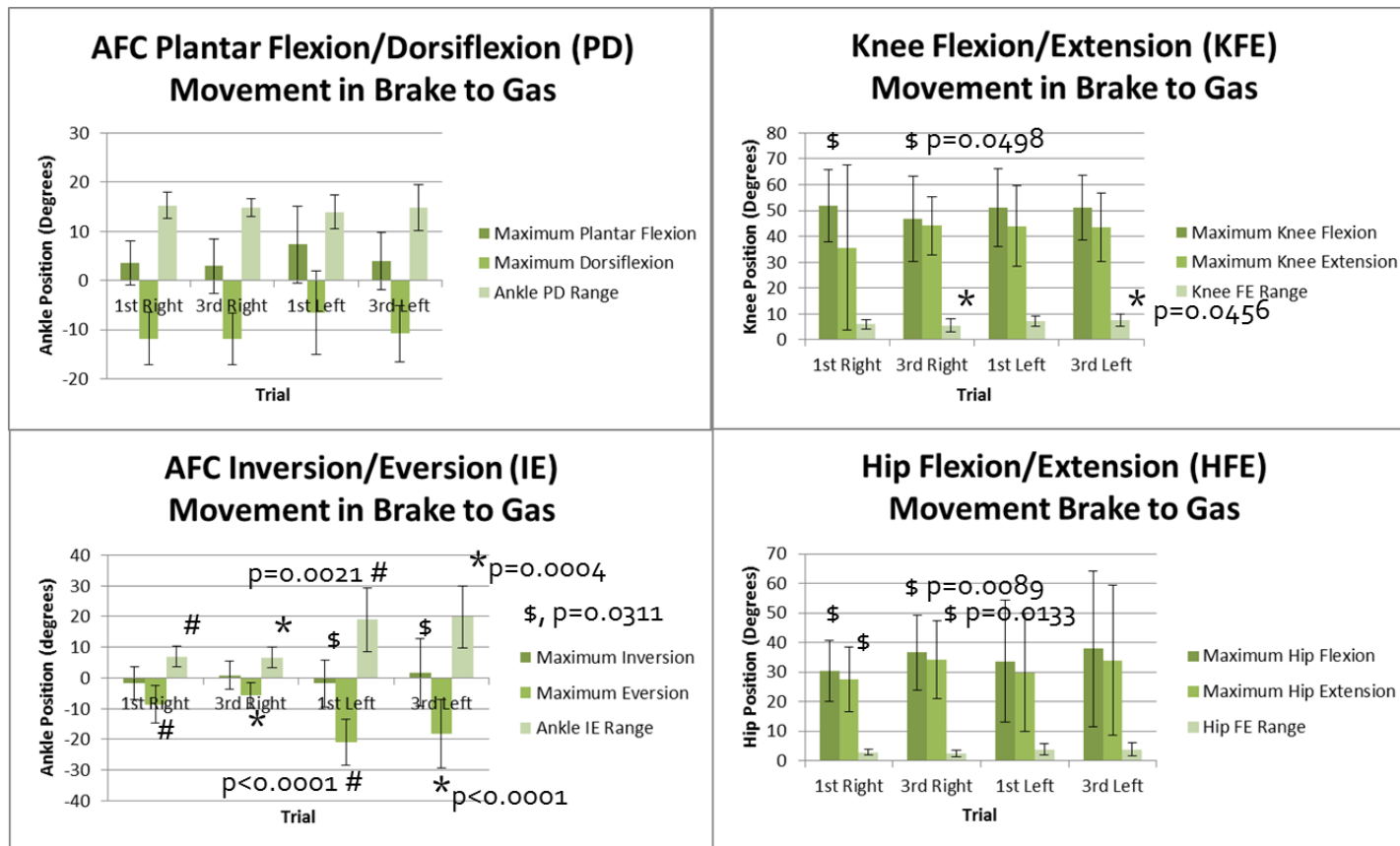


Figure 12: Joint angle motion averages for all trials in the return from brake to gas phase of the braking task. Minimum values are recorded as the physical manifestation of each movement. AFC IE minimum is Eversion, KFE minimum is Extension, HFE minimum is Extension, and AFC PD minimum is Dorsiflexion.

Comparing the first trials of the left and right leg, there are 8 variables which were found to have SSDs, six of which were found to also have SSD in the braking response movement. All eight of these variables were found to be SSD in the third trial as well. When comparing right to left, we see SSD in the peak eversion (first trial, -8.6 ± 6.2 and -20.9 ± 7.4 respectively, ($p < .0001$); third trial -5.8 ± 4.1 and -18.1 ± 11.1 respectively ($p < .0001$)) and IE range (first, 6.9 ± 3.4 and 18.9 ± 10.4 respectively ($p = 0.0021$); third, 6.6 ± 3.3 and 19.8 ± 10.0 , respectively ($p = 0.0004$)). The drastically different eversion and IE range values have an impact on many of the slope and correlation values as found in the right column of table 6. Some of these parameter effected by this increased IE motion include PD/IE slope (0.2 ± 0.2 and 0.9 ± 0.6 , respectively ($p = 0.0016$); 0.1 ± 0.2 and 1.0 ± 0.5 , respectively ($p = 0.0002$)), KFE/IE slope (-0.5 ± 0.7 and -2.0 ± 1.7 , respectively ($p = 0.0106$); -0.5 ± 0.5 and -2.3 ± 1.5 , respectively ($p = 0.0048$)), and HFE/IE correlation (0.3 ± 0.2 and $0.5 \pm .2$, respectively ($p = 0.0048$)); 0.3 ± 0.3 and 0.6 ± 0.2 , respectively ($p = 0.0041$)) and slope (-0.5 ± 0.8 and -4.3 ± 4.7 , respectively ($p = 0.0105$); -0.9 ± 1.0 and -4.7 ± 4.7 , respectively ($p = 0.0220$)). SSDs were also found when comparing parallel trials of the left and right in the variables of KFE/PD slope (-3.1 ± 1.5 and -2.1 ± 1.2 , respectively ($p = 0.0115$); -3.4 ± 1.5 and -2.1 ± 0.9 , respectively ($p = 0.0171$)) and HFE/KFE correlation (0.5 ± 0.3 and 0.7 ± 0.2 , respectively ($p = 0.0477$); 0.5 ± 0.3 0.8 ± 0.1 , respectively ($p = 0.0029$)).

In addition to the variables mentioned above, other variables found to be SSD in the third trial when comparing right to left but not in the first include KFE range (5.4 ± 2.6 and 7.4 ± 2.3 , respectively ($p = 0.0266$)), HFE/PD correlation ($0.5 \pm .3$ and 0.8 ± 0.1 ,

respectively ($p=0.0122$)), and KFE/IE correlation (0.4 ± 0.2 and 0.6 ± 0.2 , respectively ($p=0.0041$)). As with the braking phase data, the return to gas data show several instances where AFC inversion/eversion was SSD when comparing right to left leg. This is for the same reasons as the braking phase, as the driver needed to put their left leg across the center line of the body to reach the gas pedal, coming from a highly everted position on the gas pedal to a more neutral position over the gas pedal.

For return to gas motions, there were no statistically significant differences in the variables of PD maximum, minimum, or range, KFE minimum, HFE range, PD/IE correlation, KFE/PD correlation, KFE/PD slope, and HFE/KFE slope.

As shown in the charts below, the characteristic relationships seen in the braking phase do not mimic that of the return to gas phase. The left leg comparisons of different joints appear to yield much more linear relationships than the braking phase did, especially in the fields involving AFC inversion/eversion. We can see that a curl is associated with each of the inversion/eversion fields at the point of maximum inversion, which may be indicative of reaching the pedal and establishing a level application of said pedal. These charts show that the application of the gas pedal (and the brake pedal) is done so with the foot in a more neutral position when compared to the position of the foot during movement. When we look at the right foot, however, there is a trend for an initial and final location. Concentrating on the AFC IE comparisons, we see that there are two clusters of data points, with sparing data collected in between each of the points. This in between data is the true movement from one pedal to the other, while the clustering of data helps to indicate a starting and stopping position. For this scenario, the starting

position is the brake pedal, and coincides where the inversion values are more positive (right leg only, associating with AFC inversion). The final position coincides with the AFC being less inverted, and closer to neutral with a less negative AFC IE value. The relationship is harder to define, as it does not appear to have a set pattern, seeming to be circular with no definitive patten or endpoints. This may be a result of the driver having good control at those joints, capable of using each joint independently and unaffiliated with a movement at another joint further down the kinetic chain. This idea is supported because by the fact that the left leg exhibits more linearity when comparing each of the joints in the return to gas phase, and drivers have less fine motor control of these joints.

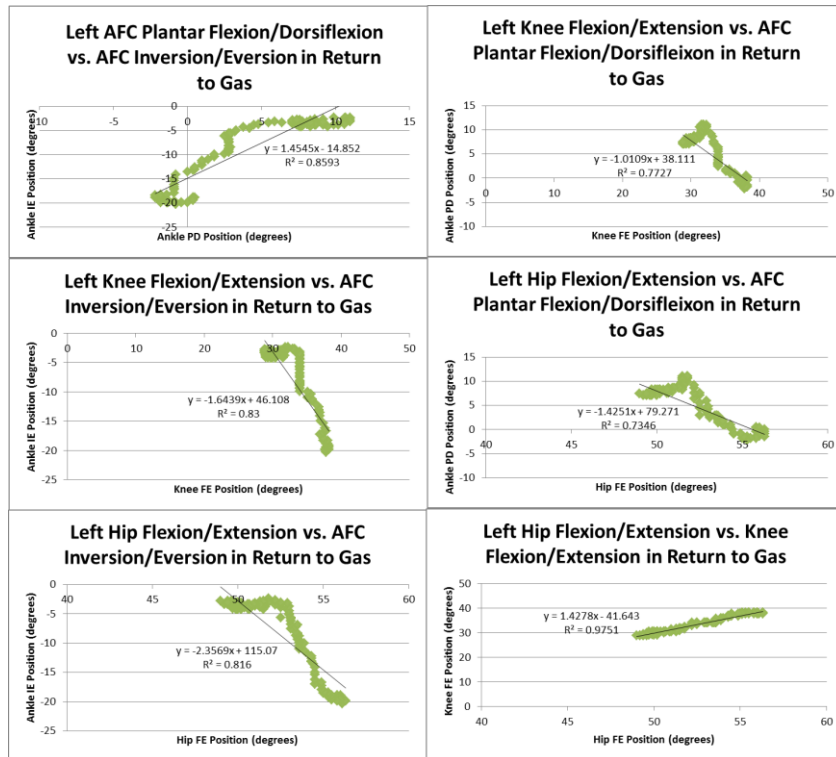


Figure 13: Collection of left leg driving motions while in the return to gas phase of braking. As seen in the figures at left, there tends to be a curl associated with AFC inversion/eversion, occurring at points of maximum inversion. This may be associated with the foot reaching the pedal and meeting resistance upon doing so.



Figure 14: Right Leg Comparisons in Brake to Gas movements. Based on the graphs, we can see that the movement in the right leg is very deliberate. There appears to be two clusters of data points in the inversion/eversion comparisons, indicating a point at which the data was first collected on the brake pedal, through the movement to the second cluster, when the foot reaches the gas pedal. The cluster with the less negative AFC IE value is the point at which the foot is on the brake, and the less negative (more neutral) position represents the foot position upon gas pedal application.

General Observations

With respect to differences between left and right footed driving, if the minimum and maximum were SSD, but the ranges were not, this indicated that the same joint range of motion was used for both the right and left legs, but the orientation or placement of the

legs was different. Essentially, the same magnitude of joint motion was required, but the leg placement necessitated that these motions used a different portion of joint motion arc.

This supports the BRT values, indicating the similar motions were undertaken, and that these motions did not increase or decrease in speed. The AFC IE range of motion was SSD between the first trials of left and right, supporting the understanding that the position of the foot between left and right footed driving requires different AFC mechanics. Left-footed driving requires a significantly greater need to perform AFC eversion to approach the brake pedal, while right-foot application of the brake pedal requires AFC inversion. This is further confirmed with the observation that with comparison of third trials of the right and left feet, these SSD in IE range are still present.

Changes from the beginning to the end of the trials in either leg are indicative of a postural conformation, an increased comfort level in the driver's seated position, or both. These changes are the numerical representation in small changes in the kinematic chain response over time. Of the joints listed in the sagittal plane, plantar flexion/dorsiflexion of the AFC was found to have the greatest range of motion (right, 17.1 ± 3.5 first and 17.0 ± 3.6 third; left, 16.2 ± 4.2 first and 18.4 ± 5.5 third) in both limbs, indicating that the AFC is the primary mover for gas and brake pedal application in this adult male sample. Knee flexion/extension showed the second greatest ROM (right, 7.2 ± 1.9 first and 7.3 ± 2.4 third; left, 8.3 ± 1.9 first and 9.1 ± 2.3 third), while the hip joint performed the least amount of movement in both legs, having a ROM of 3.5 ± 1.1 (first) and 3.6 ± 1.2 (third) in the right, and 4.5 ± 2.1 (first) and 4.8 ± 2.5 (third) in the left.

The range of motion measure is an important variable to consider, as minimums, maximum, and slopes may change as a result of a change in posture from trial to trial. Slope magnitudes can't measure shifts in joint ranges of motion, nor can they measure changes in the magnitudes of these ranges of motion, so it is important to make specific observations of absolute values on Min, Max and Range, as was done above. The incorporation of all these data is what gives the best picture of the kinematic chain while operating the gas and brake pedals in a motor vehicle. In general, we see the strongest slope values in the right leg associated with those involving AFC plantar flexion/dorsiflexion, reinforcing the range of motion observations made above. The AFC was found to be the primary mover of gas and brake application in both the right and left, followed by the knee, and then the hip. With respect to the left limb, however, we actually see a nearly equivalent level of AFC eversion as plantar flexion when it comes to brake application the range of motion associated with the AFC inversion/eversion for the left leg (first and third trial) was found to be 23.9 ± 12.0 and 25.4 ± 11.9 , respectively- nearly 15 degrees greater than the corresponding right AFC inversion/eversion ROM (9.9 ± 3.6 and 9.5 ± 4.0 , respectively). The increased joint motion at the secondary axis of the AFC is also found in the return from brake to gas phase of braking, with the left AFC showing a range of motion of 18.9 ± 10.4 in the first and 19.8 ± 10.0 in the third, and SS less motion at the right AFC (6.9 ± 3.4 first ($p=0.0021$) and 6.6 ± 3.3 third ($p=0.0004$)). This profound difference in the AFC joint range of motion indicates that the kinematic chains of the left and right leg are similar, but have some very distinct characteristics.

DISCUSSION

Over the course of testing, the Left Leg showed more SSD changes in the magnitudes of MAX and MIN values than the right leg (4 vs 1, respectively). Analysis of the correlation values indicates that in general, left leg correlation values were consistently higher than that of right correlation values, showing this trend in trials 1 and 3. Although there was not much change in kinematic chain slopes or correlations from first to third trial in either leg (HFE/PD correlation in the left leg of both braking and return to gas was the only correlation or slope that was SSD (greater) in the third trial when compared to the first), the difference in the measures of those R squared correlations may be indicative of driver behavior overall. This measure could be interpreted as the "confidence" of the motion over the course of the trial. Drivers are assumed to be more confident while performing the task with the right leg when compared to the left leg, and have a lower correlation value associated with the joint motion relationships. These lower correlation values- all but one joint showed this to be SS lower (KFE/PD)- are an indicator of fine motor control, and the lack of a need to perform the task the same way every time. With a lack of confidence in the pedal operation of the left leg, there is an increased need to perform the task in the exact same way every time using gross motor movements.

The slope values were typically found to be greater in left leg driving, with the exception of the KFE/PD movement. Each of these slopes has a very specific meaning, helping to characterize the motion in terms of direction and proportion for any given movement. For instance, the HFE/PD slope in trial 3 of the left braking response was -

4.7±4.1. This means that for every one degree of extension by the hip, 4.7 degrees of plantar flexion was performed. Another example, HFE/KFE for the same trial was found to be 2.0±1.7, or 2 degrees of knee extension for every one degree of hip extension. This trend continues for all of the slopes, with negative values being associated with dorsiflexion, eversion of the right and inversion of the left, and hip and knee extension, while positive values are associated with plantar flexion, inversion of the right and eversion of the left, and hip and knee flexion. The inversion/eversion sign notation is different from the others because of the instrumentation technique. In order to normalize the min, max, and range values of all the other joints, sign notation had to be switched on the left so it mimicked the right, with the exception of the secondary axis at the AFC. This switch caused the slopes to be inverted as to what they should be. An example of a slope involving inversion/eversion is KFE/IE, with the third right trial having a slope of -0.3±0.7 and the left having a slope of -2.8±1.8. The right trial slope means that for 0.3 degrees of *inversion*, there is one degree of knee extension associated with it, while the left trial slope indicates that for every 2.8 degrees of *eversion* there is one degree of knee extension associated with it.

Taking all of the variables into account, there is a trend showing that AFC PD has the most motion, followed by the knee, then the hip AFC IE varies from right to left, involving the least amount of motion relative to other joints in the right, while performing more motion- or at least equal to AFC PD- in the left limb.

The “minimum” values of each of the joints, also known as points of maximum eversion, dorsiflexion, and hip and knee extension only had noticeable differences when

looking at the AFC eversion. This is because of the pedal setup, requiring the driver to perform more movement at AFC joint in the transverse plane. The eversion of the left AFC is very pronounced in both the braking and release aspects of the braking response. Eversion is not expected to decrease in subsequent trials, as the movement from the gas pedal to the brake pedal (as performed by the left foot) requires this motion to depress the pedal accurately. Outside of the AFC eversion, no statistically significant differences were found when comparing the values of left vs. right trials in joint angle minimums. The only instances where SSD were found was an increase right hip extension from first to third trial, in both the braking and return to gas phase. This means that the hip was more involved in the braking task (inherently starting at a greater extension before returning to the gas). AFC dorsiflexion is limited simply because of the instrumentation and standardization of joint angles. The test subjects were zeroed at anatomical position, which means the AFC is already bent at 90 degrees. Most people don't have an extended range of motion with respect to AFC dorsiflexion, so very little motion is done at the AFC in order to move the foot off one pedal or another. With respect to knee extension, we wouldn't expect there to be a SSD between the legs, as it coincide with the application of the brake pedal, which has a more finite limit to the amount of pressing that can occur.

The maximum values, or points of greatest inversion, AFC plantar flexion, and hip and knee flexion, show a trend of the left leg having a higher maximum value. There were no joints which showed a SSD between the maximum values of right and left, though maximum AFC plantar flexion was found to be higher in the left leg in both

braking trials 1 and 3 (right, 4.8 ± 5.1 first and 4.5 ± 6.0 third; left, 8.0 ± 7.3 first and 5.4 ± 6.2 third) and return to gas (right, 3.6 ± 4.5 first and 2.9 ± 5.5 third; left, 7.3 ± 7.8 first and 4.0 ± 5.8 third). The maximum values of the knee and hip coincide with the point at which the leg is as far off the brake or gas pedal as possible. AFC inversion appears to be a non-factor for both of the limbs, having maximums near 1 degree of inversion. This means that as the right leg depresses the brake pedal (and thus their starting position for the return to gas), very little AFC inversion is occurring. This may be due to the fact that the pedals are better centered over the right foot, or possibly that the motion performed to accomplish this rotation to and from the brake pedal was not captured by our means of kinematic assessment. The increase in left AFC plantar flexion movement when compared to the right AFC could be an indicator of a lack of fine motor control, unable to have a smaller motion of tapping the brakes, as opposed to slamming on the brakes, giving the driver and their passengers the experience of an unsmooth ride. Knee and hip flexion are amplified in left leg as well, having the need to pick the foot higher off the pedal to accomplish the movement from one pedal to the other.

The range of motion data was found to be greater, in some instances SS greater, in the left leg. While not all of these were found to be of statistical significance, the trend of an increased range of motion seen at the left limb indicates that more dynamic motion was performed by the left leg while operating the gas and brake pedals. This may be a quantitative manifestation of the lack of fine motor control in the left leg. The smaller ROMs could be learned over time in right leg driving, or could be the result of having the proprioceptive knowledge of the pedal well, or knowing exactly where the pedals are

with respect to the leg without visual cues. This same kind of proprioceptive feedback might not as refined- though prior knowledge of the pedal well give some insight- in the left leg, resulting in these larger motions both to and from the brake pedal.

AFC inversion/eversion measures are where the majority of the differences were found in this study. In terms of numbers between braking and return to gas phases, of the 50 instances where SSD exist, 30 of those involved AFC inversion/eversion- more than any other joint. We see the left leg showing much more motion in this secondary axis when compared to the right leg. Across the trials in both phases, peak inversion shows no SSD between right and left, while the values of greatest eversion show SSD, making the ranges also SSD. These motions which differ greatly from the right leg have a significant impact on the slope values associated with left leg driving, showing SSD in all three associated slopes (KFE/IE, HFE/IE, and PD/IE).

The three errors described in the “Laboratory Process and Statistical Analysis” section of this paper have distinct characteristics that differentiate themselves. In total, there were 524 total errors as defined by the authors across the 12 test subjects. This first figure (Figure 15) is what is regarded as a typical braking response. There is a steady level of gas pedal application maintaining the required speed, and then the brake stimulus (dashed black line) is triggered, and the test subject responds by releasing the gas pedal without depressing it further, and moves to hit the brake pedal. The foot then recovers and moves back to the gas pedal, where the level of depression is similar to that of the level prior to braking. The solid black line represents the maximum level of gas pedal depression for each gas phase, or the time in between braking tasks. The ‘y’ axis of these

graphs are represent the level of depression as a percentage, as measured by a potentiometer, and the 'x' axis is the time associated with each trial.

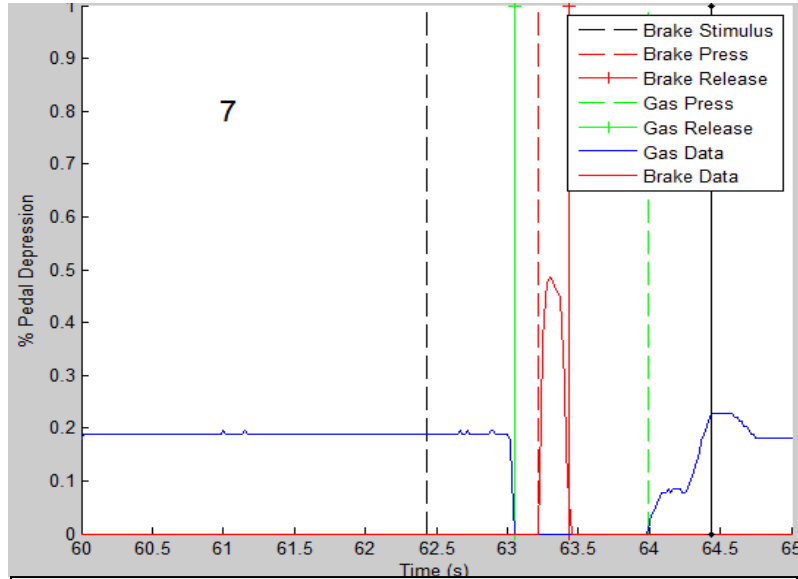


Figure 15: A typical braking response. Note that there is no increase in gas application after braking stimulus (dashed black line). Gas data is shown in blue, while brake is in red. The gas release is notated by the solid green line, braking application by the dashed red line and release by the vertical solid red line, gas pedal onset by the dashed green line, and peak gas application during any gassing phase by the vertical solid black line. Vertical Axis is the level of depression as a percentage, as measured by a potentiometer.

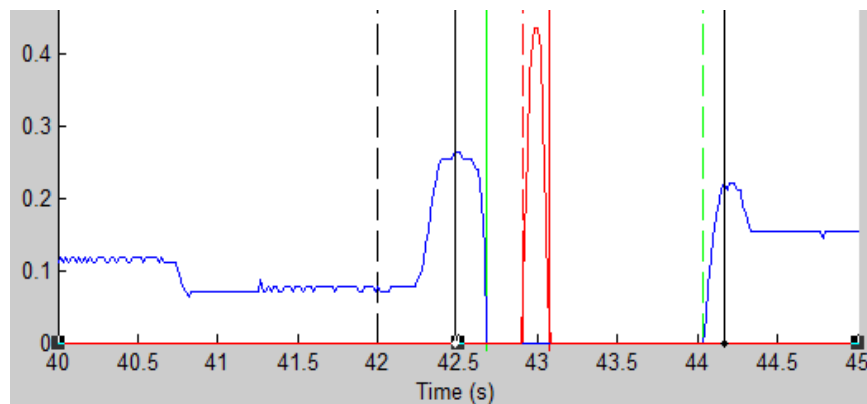


Figure 16: Type "A" Error. The gas pedal application increased by greater than 5% (about 20% here) before a correct braking response occurs.

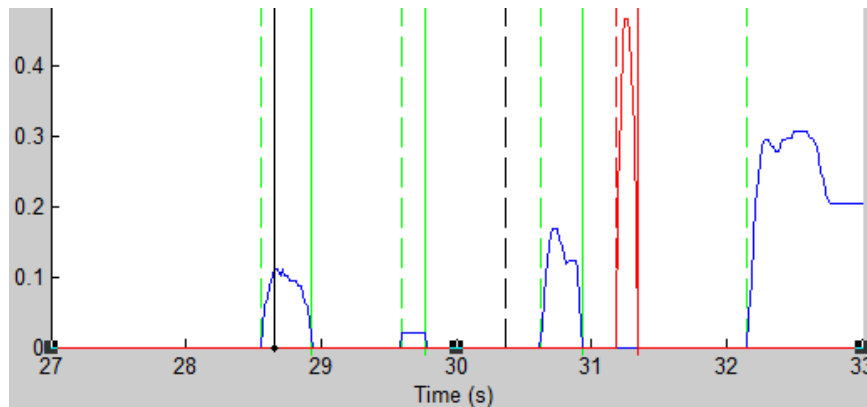


Figure 17: Type “B” Error, an independent gas peak occurs after a brake stimulus has occurred. The multiple peaks prior to the braking stimulus exist because subjects were required to maintain a speed, rather than depress the gas pedal at all times.

The type “A” error (figure 16), which is the notation given for braking responses which first elicit an increased gas pedal application before moving the foot to the brake pedal. This error was by far the most prevalent of the errors characterized by the authors, with 413 of the total 524 errors being this type “A” error. The figure at right is representative of this type of error.

The type “B” error refers to instances where the gas peak is depressed independently of the gas phase, with the pedal application initiating after the brake stimulus has occurred. Such an error would manifest on the road as a rapid acceleration when braking is intended. These peaks were typically very sharp increases, followed by an immediate release and then the correct braking response. The figure shown (figure 17) is an example of what this kind of response may look like. The reason there are multiple independent peaks in the gas phase is because the driving task required the driver to maintain a certain speed. This means that the foot doesn’t have to be on the gas

pedal the entire time, and coasting is possible at any time, as shown by the absence of gas or braking data at certain time points. Again, the dashed black line is the braking stimulus, with the error being the peak starting after the dashed line immediately before the braking response, shown in red. Of the 524 errors, 86 were this kind of independent gas peak error.

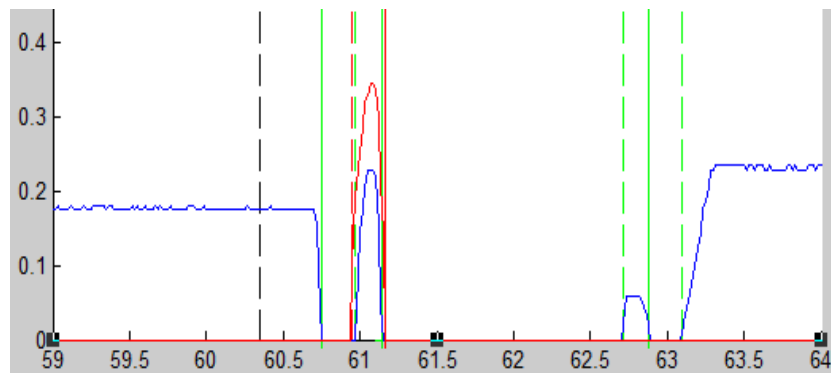


Figure 18: Type “C” Error. Both the gas and brake pedal are depressed simultaneously. The brake pedal is depressed to a greater extent because the brake pedal is raised in comparison to the gas pedal, just as in a real car.

Finally, the type “C” error is characterized by a simultaneous depression of the gas and brake. These errors occurred the fewest times, with 25 of the 524 errors. The error shows in the graphical representation as the brake pedal being depressed to a greater degree than the gas pedal, and this is due to the fact that the brake pedal is slightly raised higher in the pedal well when compared to the gas pedal- as with all automobiles. A typical “double depression” error is seen here (figure 18).

"A" Left Errors	"A" Right Errors	"B" Left Errors	"B" Right Errors	"C" Left Errors	"C" Right Errors
95	99	71	77	88	94

Table 8: Trial by Trial breakdown of the quantity of errors

Perhaps one of the most interesting aspects of these errors is the quantity of the errors from trial to trial. Of the 524 errors, 270 were committed by the right foot, and 254 were committed by the left foot. This is not to say that the left foot is “better” than the right foot, simply that fewer errors (as defined by the authors) were committed by the left foot. The trial by trial breakdown of these errors was as follows:

Clearly, there is no trend from beginning to end when it comes to the errors committed by each foot. The fewest number of errors occur in the second trial with each leg. One possible reason for this may be that the first trial is still uncomfortable for the driver despite the training scenarios. The second trial may have manifested as the trial where the drivers were still “fresh”, but now more comfortable with the driving scenario performed. The third trial may show an increase in errors simply due to the fact that it was the final segment of work at the end of an hour and a half session in a driving simulator, and fatigue may have set in. Additional studies need to be performed to assess if the increased error response is due to fatigue, or to some other reason.

Though the errors did not show a significant trend, the left leg committed 16 fewer errors. This may have been impacted by the fact that the reaction time was slower, and the motion of the left limb was more pronounced. This greater motion takes a longer time to accomplish, but this more dynamic response may reduce the risk of pedal errors, by decreasing the likelihood of an independent gas peak or simultaneous gas and brake depression.

The test subjects used in this study were of similar background, having been regular drivers in the United States, of similar stature, age and shoe size, and all were

healthy for all intents and purposes of this study. It is difficult to say with any certainty if the subjects used in this study had any kind of significant impact on the outcomes, when compared to different demographics and age groups. The drivers selected for this study had adequate experience driving in the United States, so it is unlikely that experience discrepancies had an effect on this study.

Study Limitations and Recommendations for Future Work

There were several limitations in this study that should be noted and improved upon in future work. While the Biometrics system is a very cost-effective method of analyzing the motion at the lower extremities, the exact location of the limbs with respect to the pedal well and the gas and brake pedals is only partially completed. Implementing some form of visual conformation would prove to show more clearly what happens with each of the errors and why, as well as what the movements look like.

One notable improvement that could be made to the experimental setup is a better synchronization technique between the Biometrics and driving simulator systems. This study synchronized the two systems by pressing the “start” button simultaneously, leaving some lag issues to human error. If the systems can be started simultaneously with one button click and initialized at the same speed, the correlations could be more accurate to the landmarks provided by the driving simulator.

Most importantly, several studies can be segmented off of this seminal work. This study went straight into a demanding, multi-variable driving simulator scenario in an area that has not had much previous work done. Assessing the movements in a more controlled environment in a simpler scenario may shed some light on how the leg

movements are different in the most rudimentary sense, without having to deal with variables such as peripheral scanning and speed maintenance.

In order to truly track learning, more than 3 trials should be conducted, and ideally, it should occur over several weeks to see if the learning actually sticks. Tracking left-leg driving as it changes from the first session to the fifth session five weeks later, for instance, could help establish any kind of learning had occurred. Conducting three trials in fairly rapid succession won't truly establish learning, but will more so establish a level of comfort not had prior to the study, which could open the door to the development of a kinematic chain with significantly improved fine motor movements, and a more fluid driving action.

Although it was not a considered variable in the current study, the results described in the current work could be affected by the degree of osteoarthritis (OA) in the patient, which plays a significant role in the effects of aging, inhibiting and limiting the mobility of elderly people because of pain associated with joint motion²⁴. Manual muscle testing experiments in the past have shown that muscle function is limited by the degeneration of bone. This is likely due to the fact that OA is painful in the joints, and these muscle contractions exacerbate that pain²⁴. The resulting inactivity from OA results in a more sedentary lifestyle, causing decreased muscle mass. In a study performed by Diracoglu et. al., they found that where clinical assessments of muscle function didn't find performance shortcomings, MMT can detect where the muscles are lacking in functional strength²⁴.

There are a few recommendations for future methodological improvements in this work. First, it would have been good to know the starting location of the foot before each braking event. Ideally, the foot starts at the gas pedal, and then moves to the brake when a brake stimulus occurs. However, in this study, test subjects were asked to maintain a speed of 55 mph, which meant that there were several instances where the foot was off of the gas pedal and the vehicle was coasting when a brake stimulus occurs. Ensuring the foot position immediately prior to the brake stimulus would go a long way in giving more weight to the joint correlation values.

Future studies mimicking this experimental setup should include the testing of elderly drivers, ages 65 and older, and comparing them to that of younger drivers. However, these test subjects should only be tested at the right leg, and not the left leg. This is so the characterization of daily driving motions for elderly drivers may be completed and compared to what would be seen in typical, healthy drivers. The ideal outcome of such a study would be to show that young drivers' movements are not significantly different from each other, but do show a significant difference to the performance of elderly drivers. The pedal application errors should also present some differences, with elderly drivers being more likely to commit one of the 4 pedal application errors described in the laboratory analysis portion of this paper.

Eventually, this technology and Biometrics setup could be employed in an instrumented vehicle. Such a vehicle should be capable of recording accelerations, braking, and turning, along with many other driving variables. The driver assessment could then be analyzed in such a way that it incorporates not only car motion, but also

driver motion. Driver motion will be heavily influenced by the accuracy and precision with which the kinematic chain is operating, and the comparison of young vs. old drivers should give some insight as to what the degradation of the kinematic chain may look like.

Driving, like walking, involves motion on the whole-leg scale. The proportion and timing of these individual joint movements makes up the kinematic chain, and the kinematic chain can only be characterized by individual joint motion capture and understanding the joint correlations associated with the motion. This study begins the process in understanding the lower extremity kinematic chain with respect to driving.

CONCLUSIONS

This study has established a new method in the assessment of lower extremity mobility in driving. To date, very little work has been done to attempt to characterize the movements of the legs during driving. The scope of current literature is limited to the assessment of pedal misapplication, and to quantify brake reaction time, as well as assess how various central nervous system defects affect driving. However, pedal misapplication studies have only been observational- looking back at police reports to identify what the driver believes caused the accident, not assessing the driver's movements. Brake reaction timers have been a part of the assessment of drivers for the better part of three decades, but these timers don't assess how these braking motions are occurring. This study has set a precedent that driving can in fact be characterized by a series of motion tracking sensors, so as to establish movement patterns not seen or studied before. With future work, this technique of quantifying driver motion can be

employed to help individuals with orthopaedic issues recover from injury or surgery, and allow them to regain their independence.

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