CALIFORNIA STATE UNIVERSITY, NORTHridge

USING OPENSIM TO DETERMINE MUSCLE CONTRIBUTIONS TO FRONTAL PLANE PELVIC MOTION

A thesis submitted in partial fulfillment of the requirements
For the degree of Master of Science in Kinesiology

by

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ABSTRACT

USING OPENSIM TO DETERMINE MUSCLE CONTRIBUTIONS TO FRONTAL PLANE PELVIC MOTION

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Low back pain is the number one musculoskeletal complaint by people eighteen years and older. While there are several causes of low back pain, poor pelvic control in the sagittal or transverse plane has been identified as one of them. Less is known about control in the frontal plane (pelvis list). Although we know the effects of some muscle groups on pelvis list, we do not know the effects of all the main muscle groups on pelvis list, nor do we know the extent of each muscle group’s contribution to it. Determining the control of any body segment is difficult because even muscles that do not attach to the segment can accelerate it. Induced accelerations using musculoskeletal models can provide insight into how muscles control the pelvis. To date, the biomechanics laboratory at California State University, Northridge has not had the ability to create musculoskeletal models and run computer simulations off of them. With the advent of a freeware, OpenSim, funded by the National Institutes of Health, costs associated with performing computer simulations are no longer a barrier. The major difficulty is in figuring out the workflow and setting up the extensible markup language files for use in
the freeware. The purpose of this study was to: 1) develop a three-dimensional musculoskeletal model simulation of gait and pelvis list (frontal plane pelvic motion) from data collected during experimental conditions; 2) verify the model and simulation by comparing the induced acceleration results of the gait trial with previously published data; and 3) perform induced acceleration analysis on a trial emphasizing pelvis list to determine the effects of several muscle groups on pelvic motion in the frontal plane.
CHAPTER I
INTRODUCTION

Low back pain is the number one musculoskeletal complaint [1]. Although there are several causes of low back pain, poor pelvic control has been identified as one of the main contributors. While there is research on pelvic control in the sagittal and transverse planes [2, 3], there is limited research on control in the frontal plane. Frontal plane pelvic motion is very complex and even those who are very proficient at it, such as oriental dancers, use different techniques for achieving frontal plane pelvic motion: they use their knee, hip, and spine muscles.

Statement of the Problem

While we know the effects of some muscle groups on frontal plane pelvic motion (pelvis list), we do not know the effects of all the main muscle groups on pelvis list, nor do we know the extent of each muscle group’s contribution to it [4]. Determining the control of any body segment is difficult because even muscles that do not attach to the segment can accelerate it [4]. Induced acceleration analysis using musculoskeletal models can provide insight to how muscles control the pelvis.

Purpose

To date, the biomechanics laboratory at California State University, Northridge has not had the ability to create musculoskeletal models and run computer simulations. With the advent of a freeware, OpenSim, funded by the National Institutes of Health, costs associated with performing computer simulations are no longer a barrier [5]. The major difficulty is determining the workflow and setting up the extensible markup language files for use in the freeware. The purpose of this study is to: 1) develop a three-
dimensional musculoskeletal model simulation of gait and pelvis list (frontal plane pelvic motion) from data collected during experimental conditions; 2) verify the model and simulation by comparing the induced acceleration results of the gait trial with previously published data; and 3) perform induced acceleration analysis on the pelvis list trial to determine the effects of several muscle groups on pelvic motion in the frontal plane.

**Definitions**

1. Pelvic control is the ability to control the pelvis in time and space.

2. Pelvis list is the same as pelvic obliquity. A right pelvis list denotes the right iliac crest being higher than the left iliac crest.

3. Inverse dynamics uses a given motion to estimate the joint moments that produce the motion.

4. Forward dynamics uses joint moments to predict a motion.

5. Induced acceleration analysis is capturing the effects of the torque created by each muscle on the acceleration of each segment in the system.

**Theoretical Framework**

OpenSim is free, open–source software funded by the National Institutes of Health and developed by Dr. Scott Delp’s group at Stanford University. It allows users to create and exchange three-dimensional musculoskeletal simulations. This study will develop the capability of using OpenSim in the California State University, Northridge’s biomechanics laboratory. Users will be able to study movement using induced accelerations, which has not been possible before at CSUN.
Hypothesis

Research Hypothesis

Based on the review of literature in Chapter 2, the following research hypothesis was developed:

1. The main contributors of frontal plane pelvic motion will be the oblique muscles.
CHAPTER II

REVIEW OF LITERATURE

Low back pain is the number one musculoskeletal complaint by people eighteen years and older [1]. While there are several causes of low back pain, poor pelvic control has been identified as one of them [2, 3]. Determining the control of any body segment is difficult because even muscles that do not attach to the segment can accelerate it [4]. Induced accelerations using musculoskeletal models can provide insight into how muscles control the pelvis. To date, the biomechanics laboratory at California State University, Northridge has not had the ability to create musculoskeletal models and run computer simulations. With the advent of freeware, OpenSim, funded by the National Institutes of Health, costs associated with performing computer simulations are no longer a barrier [5]. The major difficulty is determining the workflow and setting up the extensible markup language files for use in the freeware. The purpose of this study is to: 1) develop a three-dimensional musculoskeletal model simulation of gait and pelvis list (frontal plane pelvic motion) from data collected during experimental conditions; 2) verify the model and simulation by comparing the induced acceleration results of the gait trial with previously published data; and 3) perform induced acceleration analysis on a trial emphasizing pelvis list to determine the effects of several muscle groups on pelvic motion in the frontal plane.

Low Back Pain

According to 2010 data collected by the Center for Disease Control and Prevention, low back pain is more prevalent in middle-aged people and it affects more women than men [1]. It is a common reason for doctor visits and work disability. In a
nationwide study, 42% of workers complained of back pain [6]. Direct health care costs associated with back pain in the United States more than 90.6 billion dollars a year [7, 8]. Indirect costs for low back pain, which include lost wages, average to 20 billion dollars a year [7]. Low back pain is not only costly, but it is debilitating. Workers who have pain exacerbations report limitation in their daily activities, accounting for 71.6% of indirect costs for back pain [6].

People with low back pain have been shown to have poor pelvic control [2,3], an inability to control their pelvis in time and space. Because pelvic motion affects the lumbar spine, aberrant movement of the pelvis could result in excessive stress on the lumbar spine. For example, in the sagittal plane people with low back pain lean their upper spine forward to stand up from a seated position, but their pelvis stays in a posterior tilt, reducing the natural curve of the lumbar spine [2]. As a result, they place excessive stress on their low back. Healthy individuals can reposition their pelvis and get better spinal alignment when they stand up from a seated position. In the transverse plane, people with low back pain display reduced pelvis-trunk variability during gait [3]. This reduced variability is referred to as in-gait or stiff gait. Patients with low back pain are unable to move their pelvis in opposition of their trunk, which may be a protective mechanism for not creating excessive torques in their lumbar spine [3, 9]. However, muscle-tendon complexes can absorb more energy than bones, and walking stiffly may cause more pain to these individuals because bones will absorb more of the energy than muscles.

Low back pain is a health concern that needs to be further explored so that better treatment and preventative programs can be developed. With spinal mechanics, Stuart
McGill suggests taking the automaker Honda’s approach when it comes to studying movement [10]. Honda applies the lessons learned from engineering their Formula One racecars and applies them to their consumer cars. To help those with poor pelvic control, we need to study from those who display high levels of pelvic control so that we can apply what we learn from them to those who lack it. While there is substantial research on pelvic control in the sagittal and transverse planes of motion, there is limited research on pelvic control in the frontal plane.

**Frontal Plane Pelvic Motion**

Frontal plane pelvic motion is defined as pelvis list or pelvic obliquity. A right pelvis list is when the right iliac crest is higher than the left iliac crest and a left pelvis list is when the left iliac crest is higher than the right iliac crest. Pelvis list is a very complex motion and even those who are proficient at it, such as oriental dancers, claim to accomplish it through these techniques: right lateral trunk flexors are used to do a right pelvis list, right knee extensors are used to do a right pelvis list, and right hip abductors are used to do a right and left pelvis list. The technique for using the right hip abductors for a right pelvis list contradicts what we know the role of the hip abductors to be from the anatomical position. However, when oriental dancers do a pelvis list, they start in a bent knee position and allow their knees to move during the motion. When doing a pelvis list from the anatomical position, the knees are fully extended and do not move. Most people use the term hip abduction and adduction to describe pelvis list, but it is important to note that hip abduction and adduction can occur without pelvis list; pelvis list can occur as a result of either hip movement, but they are two separate movements.

While we know the effects of some muscle groups on pelvis list, we do not know
the effects of all the muscle groups on pelvis list, nor do we know the extent of each muscle group’s contribution to it. It is also unclear if the role of the hip abductors changes when the knees are allowed to move. To determine this, muscle forces of all muscle groups have to be measured or estimated. Once muscle forces are known, then the effects of different muscle groups on pelvic obliquity can be predicted.

**Measuring Muscle Forces**

*I invasive/Direct Techniques:*

Muscle-tendon force can be measured either directly or indirectly. One direct technique that has been developed to measure tendon force is using a buckle transducer, a buckle frame with a crossbar and strain gauges. This procedure requires a subject to undergo surgery in which the transducer is implanted in the body. Although local anesthesia is used during the procedure, evidence suggests that anesthesia should not be directly injected into the muscle or tendon, so that normal proprioception is maintained during the activity being studied [11]. The buckle size is matched with the size of the tendon and the crossbar is fixed under the tendon [12] so that the tendon goes through the buckle like a belt. Wires from the strain gauges are threaded through the skin and sterilized. The transducer cable is then connected to an amplifying unit.

A buckle transducer is a good way to measure continuous muscle-tendon force. However, there are many limitations to using a buckle transducer. Due to the size of the transducer, it cannot be used for measuring forces of most the muscles that control the pelvis; studies using this device are limited to the Achilles tendon [12, 13, 14]. Careful attention to calibrating the transducer and implanting it are essential for reducing error [12]. There is always a risk of infection. Pain and discomfort experienced during the
procedure and after the removal of the transducer discourage people from being subjects in studies that involve a buckle transducer.

Another direct technique used to calculate muscle-tendon force is the utilization of fiber optics. A fiber optic procedure involves inserting a fiber optic probe into a tendon using a 19 gauge (1.0 x 40 mm) needle [15] and data from the optical fiber is collected via light pulses. Implanting a fiber optic in a tendon is less invasive and it can be used on tendons besides the Achilles tendon because of its small size. However, skin contact with the fiber optic produces high measurements errors; measurement errors are reduced when using cadavers because one can remove the skin [16]. Its instability and inconsistency [16, 17] are not appropriate for use on live people, especially during dynamic trials. Additionally, their output is limited to a small sensing area, which makes fiber optics unsuitable for measuring muscle force on large muscles [18].

**Less Invasive/Indirect Techniques:**

Measuring muscle-tendon force in experiments is difficult to establish. Torque is dependent on the amount of force applied, angle of the force applied and the length of the moment arm. In experiments, only the total torque produced at a joint can be quantified. Individual muscle-tendon torque cannot be isolated, which makes establishing a cause and effect relationship challenging. This problem is further complicated by the fact that muscles can cause accelerations at joints that they do not cross [4].

One method used in experimental studies is electromyography (EMG). EMG records action potentials, electrical activity, associated with muscle activation via electrodes [19]. EMG is beneficial for determining if a muscle is active during a movement [19], but relating muscle activity with muscle force is problematic. The EMG
signal can vary depending on the type of muscle contraction. There seems to be a linear relationship between EMG and force during isometric contractions, but not during isotonic contractions [20]. It is also important to note that an EMG signal will generally be less during an eccentric movement than a concentric movement because during eccentric movements one can preload tension in tendons. There is less contribution from muscles, thus lower EMG amplitudes are recorded [21, 22].

There are two kinds of EMG electrodes: surface EMG, which uses electrical nodes to capture muscle activity, and intramuscular EMG, which inserts a needle electrode into the muscle to capture muscle activity [23]. Surface EMG is not invasive and can be performed outside a clinical setting. A couple of limitations of surface EMG are that it can record the activity of only superficial muscles and crosstalk. It can produce faulty results because it may pick up signals from nearby muscles. Intramuscular EMG can be inserted into deeper muscles and it does a better job at minimizing cross talk. It can also be used on smaller muscles because of its size. While intramuscular EMG can measure the force of the motor unit it is recording, it cannot measure all the motor units firing within the muscle. The disadvantages of intramuscular electrodes are that they are invasive, cause discomfort, and are difficult to repeat using the same area. An EMG signal does not reflect the firing frequency or the number of motor units involved with a particular movement [24] because it cannot distinguish the effects of firing frequency and number of motor units. Thus, total muscle torque cannot be determined using EMG techniques because EMG only measures direct electrical muscle activity.

A better approach for determining mechanical demands in experimental studies is using inverse dynamics. Inverse dynamics uses kinematics or kinematics with kinetics to
estimate joint torques. Cameras are usually used to capture movements via reflective markers. Once the kinematics are known, joint torques can be approximated. The most common way of determining inverse dynamics is using the Newton-Euler method, in which calculating joint reaction forces is based on the sum of the forces equaling the product of mass and linear acceleration; calculating muscle torque is based on the sum of the torques equaling the product of mass moment of inertia and angular acceleration. Because the equations are based on a link-segment model, inverse dynamics can be easily resolved with open-chain motions. During closed-chain motions, forces between the limb and the object it is in contact with must be measured. A tool used for measuring external forces is a force plate, which measures ground reaction forces.

Inverse dynamics provides means for enhancing our understanding of motion, but it has several limitations. The center of mass distribution on the link-segment model is situated at one point and remains at that point during motion. The lengths of the segments and the moments of inertia do not change during movement and it assumes frictionless joints with no co-contraction of agonist and antagonist muscles. The joint torques calculated using this method are net torques. As a result, it is impossible to determine individual muscle contributions to joint torques because there is no way of differentiating between different muscles. Thus, one cannot establish a cause and effect relationship between muscle force and segment acceleration.

While inverse dynamics uses a known motion to predict the joint torques for the motion, forward dynamics can be used to predict an unknown motion from known joint torques. Forward dynamics begins with muscle activation, which leads to muscle forces to estimate joint torques [24, 25]. It starts with the neural command and then uses muscle
activation dynamics, muscle contraction dynamics, and musculoskeletal geometry to approximate displacement [25, 26]. The neural command can be estimated using an EMG or a mathematically based optimization model [25, 26]. Using EMG to establish the neural command is difficult due to its high signal variability, especially during non-isometric contractions. The EMG method is also problematic because the process of going from muscle activation to muscle forces is not well understood; most models that use this method are based on A.V. Hill’s model [27]. An optimization approach bypasses the problems of the EMG method by directly estimating muscle forces; it does not take into account how muscles are activated.

Estimations of joint torques using forward dynamics are susceptible to error because there is not a good way to verify the predicted individual muscle forces. Muscle moment arms and line of action are difficult to individualize because they require access to advanced medical tools. Small errors in moment arms and joint torques could result in large errors in joint displacement. Nevertheless, forward dynamics allows researchers to establish cause and effect relationships and study movement in ways that is not possible with other methods.

A type of forward analysis that is used to analyze the effects of individual muscles on different segments of the body is induced acceleration analysis. Induced acceleration allows one to see a snap-shot in a time of accelerations of body segments caused by one muscle [28, 29]. It freezes the simulation at a moment in time, sets all the torques to zero except the one of interest, and determines the acceleration of a particular segment as a result of that torque. While induced acceleration gives us a picture of one particular moment in time, it has greatly enhanced our understanding of segment coordination
during movements, particularly gait. For example, several studies [28, 30, 31] have found that the plantar flexor muscles are the primary contributors to the forward progression and support of gait. These findings have brought about better treatment programs for people with pathological gait.

Induced acceleration analysis can be done in OpenSim, an open-source software funded by the National Institutes of Health that can be used with other tools and software to create, analyze and exchange three-dimensional musculoskeletal models and simulations [5]. Determining induced accelerations in OpenSim first requires: 1) scaling, 2) inverse kinematics, 3) residual reduction algorithm and 4) computed muscle controls. Scaling is matching the subject’s anthropometry to the model, inverse kinematics is measuring the joint angles for the motion, residual reduction algorithm calculates the joint torques for the motion and computed muscle control estimates the muscle activation. In essence, induced acceleration in OpenSim performs inverse dynamics on experimentally collected data and then does forward dynamics with static optimization; static optimization predicts individual muscle forces by using a criterion that minimizes muscle stress cubed.

The aim of this study is to create three-dimensional musculoskeletal model simulations based off of experimentally collected data and use OpenSim to run induced accelerations to determine the effects of various muscles on frontal plane pelvic motion. Two dynamic trials will be collected: walking gait and pelvis list. The walking gait trial will be compared with previously published data to validate the musculoskeletal model and simulation workflow. Once the gait trial is tested against other studies, the simulation of pelvis list will be created and analyzed. It is hypothesized that the greatest
contributors to pelvis list will be the oblique muscles.
CHAPTER III
METHODOLOGY

Procedures

Kinematic and kinetic data were collected and used to create a three-dimensional simulation of a walking gait trial and a standing pelvis list trial. Induced acceleration analysis was done on the gait trial to validate the methods because no three-dimensional simulations of a standing pelvis trial have been done before. Once methods were validated, induced accelerations analyses based on potentials were done on the pelvis list trial.

One female oriental dance subject, 31, was used to collect one static calibration trial and two dynamic trials: gait and pelvis list. For static calibration, she was asked to stand still and her standing body posture was captured. For the walking gait trial, she was asked to walk across a force plate at a comfortable pace. For the pelvis list trial, she was asked to stand on two different force plates with her feet hip width distance apart and her knees slightly bent. She was then asked to laterally tilt her pelvis to the right, and was instructed to allow her knees to move and not keep them rigid. This described movement is the foundation of oriental dance and is commonly referred to as a “hip up.”

Experimental Data

A total of eighteen reflective markers were placed on bony landmarks on the subject to track kinematics. The bony landmarks were: right and left head of the second metatarsal head; right and left head of the fifth metatarsal head; right and left tibia apex of medial malleolus; right and left fibula apex of lateral malleolus; right and left femur lateral epicondyle; right and left femur medial epicondyle; right and left femur greater
trochanter; right and left ilium crest tubercle (iliac crest); and right and left acromion. The functional hip joint center was estimated by measuring the midpoint of the anterior superior iliac spine and the pubis and moving inferior 2-2.5 centimeters. Weight in kilograms and height in meters were also recorded.

The static calibration trial and two dynamics trials were captured using seven Vicon (Vicon, Oxford, UK) cameras and two Kistler (Kistler, Amherst, MA) force plates. Marker trajectories were collected at 240Hz and ground reaction forces were collected at 960Hz. Vicon Nexus Software was used to label the markers and pattern fill was used to fill in the marker gaps. Pipeline was then used to export a .c3d file for all the trials.

**Skeletal Modeling & Musculoskeletal Simulations**

To generate musculoskeletal simulations in OpenSim, several files are required, one of which is a motion file. Visual3D software can be used to generate a motion file, but it requires several steps. A hybrid skeletal model was first built in Visual3D using the static calibration .C3D file from Vicon Nexus. The hybrid model contained the following segments: right foot, left foot, right shank, left shank, right thigh, left thigh, pelvis and trunk. A v3d_lab segment was created. The dynamic .C3D files were then imported into Visual3D and the motions were applied to the skeletal model. Before exporting the dynamics trials as motion files, several extensible markup language (.xml) files were placed in the same file folder to be overwritten by Visual3D scaling and inverse kinematics during motion export. Once the dynamic movements were exported as OpenSim motion files, kinematics and kinetics of the motion files were created.

OpenSim’s gait2392_simbody model was used for this study. The three-dimensional model had twenty-three degrees of freedom and consisted of seventy-six
muscles. Since the model was scaled to match the subject’s anthropometry based on experimentally collected data and joint angles were found using inverse kinematics in Visual3D, the simulation was ready for the next step: residual reduction algorithm (RRA). RRA was used to calculate joint torques. RRA minimizes the effects of errors that can lead to residuals by changing the model’s center of mass to match the torques that create the ground reaction forces. Computed Muscle Control (CMC) was then used to estimate muscle excitations, activations and the forces required to produce the torques for the simulation. CMC uses a feed forward and a feedback system by solving a static optimization problem for every moment in time within a forward dynamic simulation. CMC feedbacks the kinematics and muscle activation of each moment in time into its algorithm to generate a forward simulation that best suits the experimental data without using dynamic optimization. This method is advantageous because it is not time consuming like the conventional dynamic optimization techniques, which require thousands of forward dynamics simulations. Induced acceleration analysis was then done on the gait and pelvis list simulations.

To test the accuracy of the three-dimensional simulations, simulated accelerations of the walking gait trial were qualitatively compared to accelerations calculated from forces and another study. For the pelvis list trial, induced acceleration potentials were qualitatively analyzed to determine the contributions of various muscle groups (ankle, knee, hip and spine) to a right pelvis list.
CHAPTER IV

RESULTS

Validation Trial

Induced accelerations of the mass center (COM) of the validation gait trial simulation were compared with accelerations calculated from forces in the X (Figures 1 and 2), Y (Figures 3 and 4) and Z directions (Figures 5 and 6). Results were very similar and consistent with the trend found by Zijlstra and Hoff (2003) [32]. Acceleration of the COM was in the range of -1.5 to 1 m/s*s in the X direction, .2 to .6 m/s*s in the Z direction and -1.8 to 1 m/s*s in the Y direction.

Figure 1. Acceleration determined from forces during gait trial in the X (anterior-posterior) direction. Unit for the y-axis is m/s*s.
**Figure 2.** Acceleration determined from induced accelerations analysis (IAA) during gait trial in the X (anterior-posterior) direction. Unit for the y-axis is m/s*s.

**IAA - Total X**

**Figure 3.** Acceleration determined from forces during gait trial in the Y (vertical) direction. Unit for the y-axis is m/s*s.

**Acceleration from Forces - Y**
Figure 4. Acceleration determined from induced accelerations analysis (IAA) during gait trial in the Y (vertical) direction. Unit for the y-axis is m/s*s.

IAA - Total Y

Figure 5. Acceleration determined from forces during gait trial in the Z (medial-lateral) direction. Unit for the y-axis is m/s*s.

Acceleration from Forces - Z
Figure 6. Acceleration determined from induced accelerations analysis (IAA) during gait trial in the Z (medial-lateral) direction. Unit for the y-axis is m/s*s.

**Pelvis List**

The start and peak right pelvis list positions (Figure 7) of the pelvis list trial were analyzed. Potentials of each muscle in the model were evaluated, and percent contributions based on the overall potential to pelvis list were calculated (Figure 8). The right adductors had the greatest contribution to right pelvis list both at the start of the movement (28%) and at the peak of the right pelvic list (23%). The plantar flexors displayed the greatest change in their contribution to a right pelvis list, increasing from 13% at the beginning of the movement to 22% at the peak of the right pelvic list. The left abductors showed the same percent contribution at both positions: 21%. The right trunk flexors contributed 15% at the start position and 16% at the peak position. The remaining muscle groups had a contribution of less than 10% to a right pelvis list.
Figure 7. Start (A) and peak (B) positions of the right pelvis list trial.
Figure 8. Potentials of different muscle groups from induced accelerations of a right pelvis list trial. Numbers in percent.

Potentials in Percentage for Right Pelvis List
CHAPTER V
DISCUSSION

The purpose of this study was to create three-dimensional muscle-actuated simulations of walking gait and a right pelvis list for induced acceleration analyses. The gait simulation was used to validate the model, workflow and results of induced acceleration. Some of the challenges of the methods were with the extensible markup language (.xml) files and getting a motion file that could be used in OpenSim. This required learning another software, Visual3D. Accelerations for the validation trial determined from forces were similar to accelerations from induced acceleration analysis in all directions. The results of the anterior-posterior and vertical directions of this study mirrored that of Zilstra and Hof (2003) [32]. The difference in the medial-lateral accelerations between this study and that of Zilstra and Hof (2003) is most likely due to force plate data in the medial-lateral direction.

For the right pelvis list trial, my hypothesis of the right lateral trunk flexors having the greatest contribution to a right pelvis list was rejected. The right adductors are the largest contributors both at the start and right pelvis peak positions. In contrast to what oriental dancers claim about the right hip abductors causing a right pelvis list, the findings of this study were consistent with what we know the hip abductors to do from the anatomical position. The right hip abductors accelerated a left pelvis list. However, it is important to note that the pelvis list performed in this study was done standing in place with both feet flat on the floor. The role of the hip abductors on pelvis list during different body positions and movements need to be explored.

Physical therapists tend to focus on the hip abductors when trying to improve
pelvic control. While the hip abductors showed an acceleration potential contribution of 21% at both positions that were examined, the hip adductors had a slightly higher acceleration potential contribution: 28% at the start position and 23% at the peak position. This study brings attention to the role of the hip abductors during pelvis list when there is knee movement. It also brings attention to the plantar flexor muscles that showed the highest change to their contribution to a pelvis list from the start position (13%) to the peak position (22%). Although this study was based on one subject, it highlights the need for balanced rehabilitation and training programs.
REFERENCES


