DALHOUSIE UNIVERSITY

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Background: There is emerging interest in hip abductor function during gait and its potential relationship to knee joint pathology. During gait, the hip abductor muscles are primarily responsible for generating moments of force to control frontal plane movement. The current study investigated the relationship between hip abductor muscle function and frontal plane hip moments of force during gait.

Methods: Frontal plane hip moments of force and electromyographic features of gluteus medius were measured during walking in 22 healthy individuals. Hip abductor strength, subject anthropometrics and gait velocity were recorded. Multiple regression models were used to evaluate the relationship between the anthropometric, velocity, strength and electromyographic variables and the initial and mid-stance magnitude of the hip adduction moment.

Findings: A positive relationship was found between the initial peak moment (Nm), and both body mass and gait speed (R^2=90%). Body mass (positive) and hip abductor strength (negative) explained significant levels of mid-stance magnitude variability (R^2=62.5%). Gait speed (positive) explained significant levels of variability in the normalized initial peak moment (Nm/kg) (R^2=52%). No variables were included in the normalized mid-stance moment model (P>0.05).

Interpretation: Body mass was the key factor associated with high hip adduction moments during initial and mid-stance of the gait cycle. Increased gait velocity was associated with higher initial peaks and higher muscle strength was associated with lower mid-stance magnitude of the external hip adductor moment during walking. These findings suggest that in a healthy adult population, hip abductor strength and activation were not directly related to the hip adduction moment magnitude during gait.

1. Introduction

Understanding the biomechanics of gait is important for developing a mechanical framework upon which to base clinical decisions associated with joint pathology in the lower extremity. Many studies have focused on the net external knee adduction moment as a surrogate measure for medial joint loading, reporting differences in both the initial peak and the mid-stance magnitude between asymptomatic controls and those with knee osteoarthritis (OA) (Landry et al., 2007; Newell et al., 2008). The influence of biomechanical (Andriacchi et al., 2000; Miyazaki et al., 2002; Mundermann et al., 2005; Thorp et al., 2006) and neuromuscular (Hubley-Kozey et al., 2006; Lewek et al., 2004) gait features on this moment encompass the primary study of pathomechanical factors related to the pathology of knee osteoarthritis.

More recently, frontal plane biomechanics of the hip joint during gait have been investigated as a factor influencing medial knee joint loading (Bennell et al., 2007; Chang et al., 2005; Mundermann et al., 2005). This biomechanical feature has been associated with knee OA progression and severity (Astephen et al., 2008a; Astephen et al., 2008b; Chang et al., 2005; Mundermann et al., 2005). In a longitudinal study, Chang et al., (2005) reported that an increased peak internal hip abduction moment magnitude was directly related to a reduction in the progression of medial compartment knee OA. They suggested that hip abductor muscle weakness was a primary factor responsible for the reduced net internal hip abductor moment (Chang et al., 2005). While the rationale for the present study relies primarily on the implications for knee OA, it has also been reported that hip abductor function and hip frontal plane biomechanics are related to other knee joint pathologies such as ligament injuries (Herman et al., 2008; Russell et al., 2006), anterior knee pain (Brindle et al., 2003) and iliotibial band

friction syndrome (Fredericson et al., 2000).   This relationship between hip joint mechanics and knee joint pathology provides a rationale for examining hip joint moments and hip muscle function (strength and activation). The goal is to improve our understanding of the relationship among these variables and their potential role in producing a mechanical environment that either protects or increases the risk of knee joint pathology.

Chang et al., (2005) and Mundermann et al., (2005) reasoned that the frontal plane moment of force about the hip (adduction/abduction) indirectly represents the control of the center of mass position in the frontal plane and the subsequent position of the net ground reaction force with respect to the knee joint center. They concluded that this frontal plane control was achieved through an internal hip abductor muscle moment (Chang et al., 2005; Mundermann et al., 2005). In an otherwise healthy hip joint, the passive osteoligamentous structures are not thought to influence the internal moment (Chang et al., 2005), suggesting that this moment is largely created by the hip abductor musculature. The implication of their work is that there is a direct relationship between the frontal plane moments of force at the hip joint captured during gait and the strength of the hip abductor musculature (Chang et al., 2005).

The gluteus medius is the main hip abductor and a large portion of this muscle acts in the frontal plane to stabilize the pelvis and lower leg during gait (Gottschalk et al., 1989). As well, the myoelectric patterns of gluteus medius captured during gait illustrate the role of this muscle during walking (Mickelborough et al., 2004; Winter and Yack 1987; Wootten et al., 1990). There is however; no direct evidence establishing a relationship between hip abductor muscle strength, gluteus medius muscle activation and the hip joint frontal plane moments during gait. An improved understanding of this relationship will provide a foundation on which to design and evaluate rehabilitation protocols aimed at the hip abductor musculature.

Electromyographic (EMG) studies of gluteus medius report bursts of activity occurring during both initial and mid-stance periods of the gait cycle in healthy adults (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990). The greatest burst of gluteus medius activity occurred during initial stance, suggesting that this muscle can potentially influence the initial peak frontal plane moment of force about the hip. The mid-stance activity was estimated to be approximately 50 percent of the amplitude found during initial stance (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990). This pattern of activity corresponds in time to the dip in the frontal plane moment of force that has been previously reported during mid-stance (Astephen et al., 2008a; Astephen et al., 2008b; Chang et al., 2005; Mundermann et al., 2005). An increase in the mid-stance magnitude of the hip adduction moment has been found in those with knee OA and this higher moment corresponds to a period of single leg support (Astephen et al., 2008b). In contrast, negligible activity has been reported for the gluteus medius muscle during the late stance phase of gait (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990) and presumably this muscle would have minimal effect on the frontal plane moment during late stance. As well, no differences in the hip adduction moment peak were reported between knee OA and controls during late stance (Astephen et al., 2008a), consequently this feature was not examined. The EMG studies of gluteus medius to date have not normalized their amplitudes to a maximum effort or a physiological reference (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990) therefore we do not have a standard on which to compare or make statements regarding the relative activation amplitudes of this muscle during walking. Subsequently, we do not know what level, as a percentage of a maximum effort, the gluteus medius muscle is recruited during walking. This paper focused on quantifying the activity of the gluteus medius

with respect to a physiological reference during both initial and mid-stance periods, and the corresponding features in the net external hip adduction moment.

Based on the static model presented by Chang et al., (2005), the rational for strengthening the hip abductor musculature is to directly influence the frontal plane moments of force about the hip joint during gait. They proposed that the stronger hip abductor musculature is needed to produce the high internal hip abductor moment to reduce the shift in center of mass toward the swing limb during stance. This would subsequently reduce the moment arm length of the external forces about the knee joint. This was deemed relevant to knee OA gait as this shift was thought to reduce forces across the medial knee compartment of the stance limb (Chang et al., 2005). To our knowledge the relationship among hip frontal plane moment, temporal gait characteristics, subject anthropometrics, hip abductor EMG and muscle strength characteristics, have not been investigated in healthy individuals or those with lower extremity injuries. The present study was a first step toward better understanding these associations. Thus, the purpose of this study was to determine if the variability in the characteristics of the net external hip adduction moment can be explained by the strength of the hip abductor musculature, subject anthropometrics, gait velocity and the corresponding characteristics of the gluteus medius electromyogram captured during gait in healthy individuals.

2. Methods

2.1 Subject Selection

Participants were recruited from the general community. Twenty-two subjects participated. For study inclusion, these individuals were required to be between 35 and 55 years of age, present with no lower extremity injuries within six months prior to data collection, no known degenerative joint disease in the lower extremity, and no cardiovascular/respiratory disease or neurological disorders. All participants gave written informed consent to participate in the study, which was approved by the Dalhousie Health Sciences Research Ethics Board.

2.2 Gait Analysis

Participants were introduced to the laboratory setting and objectives of the study were reviewed. A single physiotherapist, experienced in orthopaedic physiotherapy, prepared the participants for motion capture and the electromyographical analysis of eight lower extremity muscles. The tested leg was randomly selected. For the purpose of this study, only the gluteus medius has been reported. After lightly shaving the skin and cleaning with isopropyl alcohol wipes (70 percent), circular bipolar electrodes (10 mm diameter, $0.79cm^2$ surface area, 20 mm interelectrode distance) were placed on the skin in the direction of the gluteus medius muscle fibers (SENIAM 1999). A reference electrode was placed on the anterior tibia shaft. Ipsilateral and contralateral isometric hip abduction in single leg stance was performed for validation of electrode placement to minimize crosstalk (Winter et al., 1994) and to set the appropriate gains to maximize signal amplitude without saturation. At least 20 minutes elapsed before electromyographic recordings were made. Electrodes, pre-amplifiers and lead wires were further secured with adhesive tape and nylon stocking.

Lower extremity motion during gait was captured in three-dimensions at 100Hz using two optoelectronic motion analysis sensors (Optotrak™, Northern Digital Inc., Waterloo, ON, Canada). Three-dimensional ground reaction forces were collected from a single force plate (AMTI™, Advanced Mechanical Technology Incorporation, Newton, MA, USA) that was aligned to the global coordinates of the motion capture system. The myoelectric activity of gluteus medius was recorded using an AMT-8 (Bortec, Inc., Calgary, AB, Canada) eight channel EMG measurement system (Input Impedance: ~10GΩ, CMRR: 115dB at 60 Hz, Band-pass (10-1000 Hz)). A Cybex™ Isokinetic dynamometer (Lumex, NY, USA) was employed to collect the isometric torque produced by the hip abductor musculature. All signals were analog to digitally converted at 1000Hz (16bit, +/- 2V) using the analogue data capture feature of the Optotrak™ system and stored for processing.

For motion capture, infrared emitting diode (IRED) skin surface markers were affixed to the lateral aspect of the lower extremity. This marker cluster and the motion capture procedures have been previously described (Rutherford et al., 2008). Briefly, triangular sets of IRED markers were secured to the pelvis, femur, tibia and foot. Single IRED skin surface markers were placed on the lateral aspect of the shoulder, greater trochanter, lateral epicondyle of the femur and lateral malleolus. After a standing calibration trial, the digitization of eight virtual points on predefined anatomical landmarks was completed, including right and left anterior superior iliac spines, medial epicondyle of the femur, fibular head, tibial tuberosity, medial malleolus, base of the second metatarsal and center of the posterior calcaneous.

Participants were instructed to walk at a self-selected velocity along a six-meter walkway. After three familiarization trials, at least five walking trials were collected. Velocity

was monitored with a photo-electric timing component, positioned at known distances on the walkway.

2.3 Hip abductor Strength

Hip abductor strength and the myoelectric activity of gluteus medius was measured using a series of two maximal voluntary isometric contractions (MVIC). Participants were positioned in side-lying on a secure, height adjustable exercise table (Bolgla and Uhl 2007; Laheru et al., 2007). Shoulders and pelvis were positioned perpendicular to the exercise table. The upper body was secured. The dynamometer was positioned posterior to the subject. The axis of rotation was aligned with the frontal plane axis of rotation of the hip. The lever arm pad was positioned and secured on the distal femur at known distance from the hip joint center. The test leg was positioned in line with the trunk. After one practice trial, a gravity correction trial was completed (i.e. limb supported by the dynamometer with subject relaxed). Two maximum isometric hip abduction trials were collected, separated by a 60-second rest period. Participants were instructed to abduct their lower limb into the lever arm pad and hold for three seconds while minimizing movement in the sagittal and transverse plane. Standardized instructions and verbal encouragement were given.

2.4 Data Processing

All data processing was completed using custom programs written in MatLab™ version 7.1 (The Mathworks Inc., Natick, MA, USA). The processing for the three-dimensional motion

and ground reaction force data were as follows to calculate the external hip adduction moment (Costigan et al., 1992). Technical and local anatomical bone embedded coordinate systems for the pelvis, thigh, tibia and foot were derived from the skin surface markers and digitized points (Cappozzo et al., 1997). Hip joint centers were derived using a functional estimation and a least squares optimization technique (Leardini et al., 1999). Joint angles were specified through Euler rotations using standard convention for presentation (Grood and Suntay 1983). Net external adduction moments of the hip and knee were calculated using an inverse dynamics model which combines ground reaction force and moment data, kinematic positional data, limb anthropometrics and inertial properties (Clauser et al., 1969; Grood and Suntay 1983; Vaughan et al., 1999).

The myoelectric signals from gluteus medius were pre-amplified 500x and then further amplified to best utilize the dynamic range of the data collection hardware. Raw signals were corrected for gain, bias and converted to micro-volts, full wave rectified and low-pass filtered (Reverse Butterworth 6-Hz, 4th order low pass filter) (Hubley-Kozey et al., 2006). For amplitude normalization, a 100-ms moving window algorithm identified the maximal EMG amplitude for each MVIC trial (Hubley-Kozey et al., 2006). The maximum value was used to normalize the EMG data obtained from the walking trials. All raw EMG signals were visually inspected to ensure quality recordings and the signal to noise ratio was checked. Both hip adduction moments and the electromyogram were time normalized to 101 data points using a cubic spline interpolation technique, representing one complete gait cycle.

Maximum hip abduction strength was determined using a static model that included the torque generated by the weight of the leg from the gravity correction trial and the torque recorded by the dynamometer during the MVIC. A 500-ms moving window algorithm was

employed to capture the maximum torque generated over the three-second steady state

contraction. The averaged value between the two trials was recorded as the maximal torque.

2.5 Analysis

The peak magnitude of the net external hip adduction moment waveform during early (0-

20 percent of the gait cycle) and the minimum value (i.e. the dip) during mid-stance (20 to 40

percent of the gait cycle) was identified for each subject as a raw value in (Nm) and normalized

value (Nm/kg) (Fig. 1(a)). The peak magnitudes in the electromyogram of the gluteus medius

were also identified at these time signatures.

Table 1: Statistical Models

Model Number	Dependent Variable	Independent Variables
1	Hip Adduction Moment (0-20 % Gait Cycle) (Nm)	Height Mass Gait Velocity Hip abductor Strength (Nm) Gluteus Medius Activation (0-20 % Gait Cycle)
2	Hip Adduction Moment (20-40 % Gait Cycle) (Nm)	Height Mass Gait Velocity Hip abductor Strength (Nm) Gluteus Medius Activation (20-40 % Gait Cycle)
3	Hip Adduction Moment (0-20 % Gait Cycle) (Nm/kg)	Height Gait Velocity Hip abductor Strength (Nm/kg) Gluteus Medius Activation (0-20 % Gait Cycle)
4	Hip Adduction Moment (20-40 % Gait Cycle) (Nm/kg)	Height Gait Velocity Hip abductor Strength (Nm/kg) Gluteus Medius Activation (20-40 % Gait Cycle)

2.6 Statistical Analysis

Four forward stepwise regression models ($p_{in} = 0.1$, p_{out} 0.1) were employed (Table 1).

Variance inflation factors were used to assess multi-collinearity and the suitability of the

regression model was evaluated by examining the plots of the residuals against the predicted

values. The net external hip adduction moment and strength of the hip abductor musculature

were assessed in Nm (model one and model two) and normalized to body mass (model three and

four) (Table 1). Significance was determined by alpha = 0.05. All statistical analyses were computed on Minitab™ V.15 (Minitab Inc. State College, PA, USA).

3. Results

Demographics, strength of the hip abductors and gait characteristics for the 22 participants are shown (Table 2). The peak hip adduction moment occurred at approximately 15 percent of the gait cycle with an average magnitude of 1.63 Nm/kg where the dip occurred at approximately 35 percent of the gait cycle with an average magnitude of 0.80 Nm/kg (Fig. 1(a)). The EMG waveform of gluteus medius shows a peak of 70 percent MVIC that occurred at approximately eight percent of the gait cycle followed by a rapid decrease (Fig. 1(b)). During mid-stance (20 – 40 percent of the gait cycle) a second smaller peak in the waveform was found corresponding to amplitude of approximately 29 percent MVIC while minimal activity was found throughout the late stance and swing phase (Fig. 1(b)). The electromyogram and net external hip adduction moment waveform have been superimposed to illustrate the relationship between these two gait characteristics (Fig. 2). While the peak moment and peak activation of the gluteus medius did not occur at the same time during initial stance, the increase in gluteus medius activation during mid-stance corresponds to the period of the gait cycle where the dip was found in the net external hip adduction moment.

The variability in the magnitude of the external hip adduction moment during initial stance (model one) was explained by the mass and self-selected velocity of the participant (R^2=90%, $P<0.05$). Mass accounted for 80 percent of the variability ($P<0.05$) whereas gait velocity accounted for an additional 10 percent ($P<0.05$). These relationships were both

positive. Participant mass and the strength of the hip abductor musculature explained significant variability in the mid-stance magnitude of the net external hip adduction moment (R^2=62.5%, $P<0.05$) where mass explained 52 percent ($P<0.05$) and hip abductor strength explained 10.5 percent ($P<0.05$) (model two). A positive and negative relationship was found, respectively. When the external hip adduction moment and strength of the hip abductor musculature were normalized to body mass, gait velocity explained significant variability in the magnitude of the hip adduction moment during initial stance (R^2=52%, $P<0.05$) for model three (Fig. 3). The independent variables could not explain significant levels of variability in the mid-stance hip adduction moment after hip abductor strength and the mid-stance moment were normalized to body mass ($P>0.05$). Subject height and the magnitude of the gluteus medius activation did not contribute in either model.

Table 2: Demographic and Gait Characteristics of
the Sample [Mean (SD)]

Demographics	
N	22
% Female	60
Age (yrs)	46.0 (7.5)
Mass (kg)	77.2 (16.3)
Height (m)	1.70 (0.11)
Hip Abductor Strength (Nm)	104.3 (23.7)
Hip Abductor Strength (Nm/kg)	1.37 (0.27)

Gait Characteristics	
Foot Progression Angle (^{o} toe out)	5.0 (4.5)
Walking Velocity (m/s)	1.46 (0.14)
Peak hip adduction moment (0-20%) (Nm/kg)	1.63 (0.20)
Hip adduction moment (20-40%) (Nm/kg)	0.80 (0.14)
Peak EMG Gluteus Medius (0-20%) (% MVIC)	70.0 (26.6)
Peak EMG Gluteus Medius (20-40%) (%MVIC)	29.1 (15.9)

4. Discussion

This study sought to determine if strength of the hip abductor musculature was a significant factor explaining variability in characteristics of the external hip adduction moment captured during walking at self-selected speed in healthy adults. Also, it tested whether the gluteus medius muscle activation during gait explained variability in this moment. Minimal work has been accomplished in deriving a framework on which to evaluate the effects of muscle strength on the dynamics of gait (Bennell et al., 2007; Thorstensson et al., 2007) and the

interactions found among the variables in the present study provide information relevant to making informed decisions.

First, the results for all gait characteristics were typical of values and waveforms presented in the literature. Gait velocity was consistent with previous reports for healthy adults (Mundermann et al., 2008; Rutherford et al., 2008). This was deemed important to establish for comparative purposes since muscle activation patterns (Hof et al., 2002; Ivanenko et al., 2004) and lower extremity biomechanics (Landry et al., 2007) can be affected by differing gait velocities. The temporal waveform patterns of gluteus medius had a high initial peak, a lower level of activity close to mid-stance and minimal activity in late stance as previously reported for healthy adults (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990). Our results illustrate that the peak amplitude achieved for the gluteus medius during gait was 70 percent MVIC for a short burst of activity at early stance indicating that it was not required to be recruited maximally during walking. The second peak was 30 percent MVIC at mid-stance, which was slightly lower than if it had dropped to 35 percent MVIC or approximately 50 percent of the initial peak as previously reported (Hof et al., 2002; Winter and Yack 1987; Wootten et al., 1990). This subtle discrepancy may be explained as actual values were not provided in the previous studies and values for comparison were estimated from the waveforms presented. In any case, this discrepancy does not alter the interpretation of the electromyography in the current study. The net external hip adduction moment waveform was consistent with previous reports for healthy adults. Two distinct peaks existed; one during early and one during late stance. The dip amplitude during mid-stance was approximately 50 percent of the initial peak (Astephen et al., 2008a; Mundermann et al., 2005). The characteristics of interest from the moment waveforms and electromyogram were easily identified in this healthy sample.

The current results illustrate that both body mass and gait velocity explained the majority of variability in the initial stance peak magnitude of the net external hip adduction moment. Considering mass is empirically related to the resultant magnitude of the calculated moment, this finding is not surprising. As demonstrated in model three, self-selected gait velocity remained a significant contributor to the variability in the initial stance hip adduction moment after the moments were normalized to body mass (Fig. 3). To our knowledge, the positive association between walking velocity and the peak adductor moment has only been shown for the knee (Landry et al., 2007; Mundermann et al., 2004). Our results indicate that the strength of the hip abductors is not related to the peak magnitude of this moment during initial stance in healthy adults. There are two potential reasons for these findings. First, the maximum isometric strength was 84 percent of the peak hip adduction moment and the gluteus medius activation was 70 percent of MVIC during gait. This suggests that, either portions of other muscles (gluteus maximus, tensor fascia lata) or passive support structures (iliotibial band) may have provided addition contributions to the moment that were not quantified in the current study. Finally, an error in the hip joint centre estimation may alter the moment representation about the hip. While empirical methods for estimation, based on pelvic geometry were used to first estimate the hip joint center, a functional estimation was carried out based on a least squares optimization routine on all subjects in the current study. Based on previously described root mean square errors of 13mm utilizing this estimation technique (Leardini et al., 1999), we feel this provides a reasonable estimate of the hip joint centre. This study supports that while the abductor muscles contribute, the internal abductor moments are not solely influenced by the strength and activation of these muscles examined in this study.

While the temporal EMG waveform characteristics presented in this study are similar to those found in the literature (Mickelborough et al., 2004; Winter and Yack 1987; Wootten et al., 1990), the current study did not find that the initial peak activation of gluteus medius was related to the peak hip adduction moment of force during initial stance. The temporal separation that exists in peak magnitudes for these two waveforms during early stance (Fig. 2) implies that they are not directly related. Also, surface electromyographic techniques capture gross muscle volume activation and perhaps, various components of gluteus medius, tensor fascia late, gluteus minimus and maximus were activated at distinct periods of the gait cycle modifying a direct relationship. Considering we captured a high level of activation in gluteus medius, a primary abductor (70 percent of MVIC), perhaps a minor deficit in abductor strength in the presence of lower extremity pathology may begin to manifest as alterations in biomechanical profiles during gait and should be explored.

The relationship between the mid-stance magnitude of the net external hip adduction moment and the independent variables showed that body mass had a similar relationship to that found for the initial peak of the net external hip adduction moment (model 2). This dip relates to the ability of the subject to unload the lower limb joints and, as shown in figure 2, a corresponding increase gluteus medius activity is also apparent during mid-stance. This temporal synchronization, unlike what occurred during initial stance, may assist to explain the finding that hip abductor strength was related to the moment magnitude during mid-stance. Healthy individuals in this study utilized activation levels of 30 percent MVIC during mid-stance. If individuals with lower extremity pathology had to utilize greater levels of activity, perhaps the relationship between hip abduction moment and hip abduction strength would be stronger. This finding suggests that in the non-normalized form, greater absolute hip abductor strength was

related to a reduced absolute moment of force about the hip during mid-stance (i.e. more joint unloading). After accounting for the effect of body mass however (model 4); this relationship was not significant, implying that muscle strength, independent of body mass, is not related to the normalized net moment of force during walking. In fact, no independent variables were included in the model to explain the magnitude of the normalized hip adduction moment during mid-stance.

Although there are limitations that need to be considered when interpreting the results from this study, the results provide evidence that the frontal plane hip moments of force during gait are not directly influenced by strength characteristics of the hip abductors. Hip abductor muscle strength is inherently one of the many factors that may potentially influence the moments of force about the hip in the frontal plane however; gait velocity and subject mass were found to be the main contributors. The first limitation pertains to the EMG normalization. An isometric contraction of the hip abductors from a side-lying position was used to quantify muscle strength and for normalization purposes. It may be argued that a dynamic measure may be considered more appropriate, however the approach employed was standardized. We chose to utilize a Cybex™ dynamometer to quantify muscular strength and maximal activation in an attempt to reduce investigator influence on strength measures that result from hand-held devices and to improve patient control and stability. Cynn et al., (2006) found that stability during the side-lying hip abduction maneuver increased the myoelectric activity in gluteus medius. Precautions were exercised to ensure control of the lower extremity and trunk during the maximal effort contraction; however it is recognized that this position and muscle contraction type does not resemble the dynamic, role the hip abductors during gait. Given the feedback and standardization provided, we feel that participants elicited maximum or close to maximal contractions and our

highest peak EMG amplitude during walking was 70 percent MVIC.   Our waveform analysis

was delimited to discrete variables rather than statistical pattern recognition approaches designed

to show changes in both amplitude and shape characteristics of the EMG and kinetic waveforms

(Hubley-Kozey et al., 2006; Landry et al., 2007; Rutherford et al., 2008). Discrete variables

were used in the present study since the research question related to amplitudes and not temporal

changes. It also allowed direct comparisons to values reported in the literature for the initial

peaks (Astephen et al., 2008a; Chang et al., 2005; Mundermann et al., 2005).

In summary, this study showed that hip abductor strength does not explain variability in

the initial peak of the hip adduction moment waveform in healthy adults although it did explain a

small amount of variability during mid-stance. Chang et al., (2005) found that a greater internal

hip abduction moment protected against medial knee OA progression, implying that hip abductor

strengthening was a beneficial management option for knee OA. While decreases in hip abductor

strength have been found in those with knee pathology (Cichanowski et al., 2007; Fredericson et

al., 2000) the theoretical foundation on which to prescribe strengthening exercises for medial

joint knee OA based on increasing frontal plane moments is unclear. In fact, this study found

that gait velocity and body mass explain the greatest amount of variability in the hip adduction

moment in healthy adults. If increasing the external hip abduction moment (internal hip

abduction moment) is the goal, then increasing walking velocity and body mass would achieve

this goal. Clearly, the ability of these two factors to produce a protective effect on the knee lacks

logic. Considering the importance of evaluating and developing effective conservative

interventions and prevention strategies for knee injuries, the present study suggests that further

investigations are required before concluding that altering the frontal plane moments of force at

the hip can be accomplished through strengthening the hip abductors. Further investigations on

those with knee pathology may illustrate the importance of strengthening the hip abductor musculature, but the rationale may need to be better clarified.

5. Conclusion

An appropriate biomechanical and neuromuscular framework to evaluate clinical interventions targeting musculoskeletal health is important for developing effective treatment of individuals with lower extremity pathology. Investigating the relationship between dynamic factors of gait, and subject specific characteristics provides such a model. This study found that mass and walking velocity were the key factors related to the initial peak hip adductor moment and that mass and hip abductor strength were related to the mid-stance magnitude of that moment during gait. When the adduction moment and strength values were normalized to body mass only walking velocity contributed to the initial peak hip adduction moment. In conclusion, hip abduction muscle strength and function are not key contributors to the initial peak or mid-stance magnitude of the external hip adductor moment in healthy individuals.
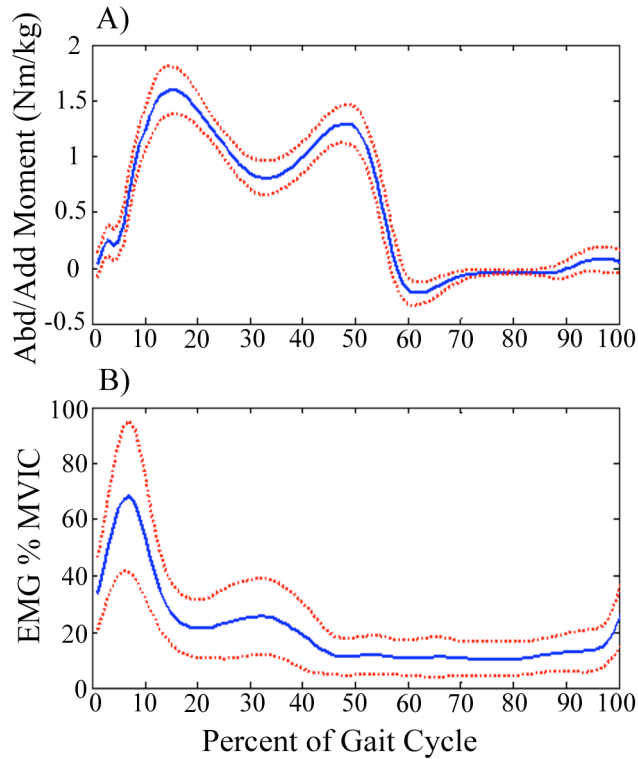
Figure 1: A) Net external hip adduction moment waveform, amplitude normalized to body mass ± one standard deviation and B) myoelectric waveform amplitude, normalized to a percentage of MVIC for gluteus medius ± one standard deviation. Ensemble average waveforms are time normalized to one gait cycle.
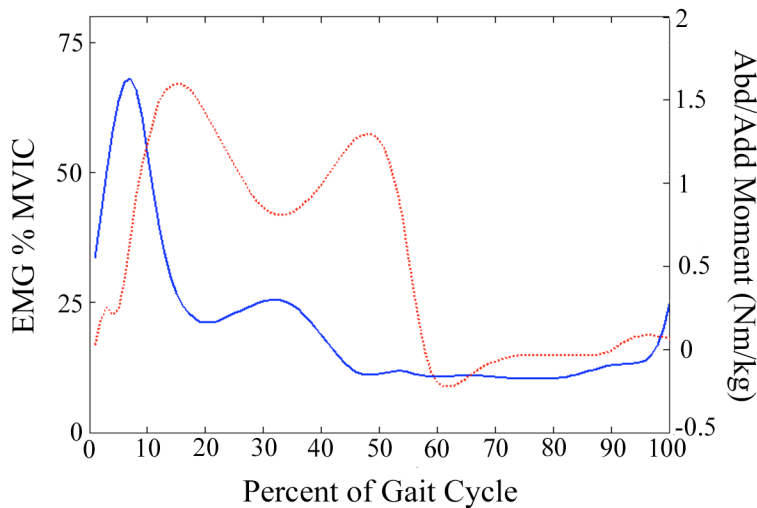

Figure 2: Ensemble average gluteus medius EMG waveform during the gait cycle shown (solid) and represented on the left y-axis superimposed onto the ensemble averaged net external hip adduction moment waveform (dotted) represented on the right y-axis.
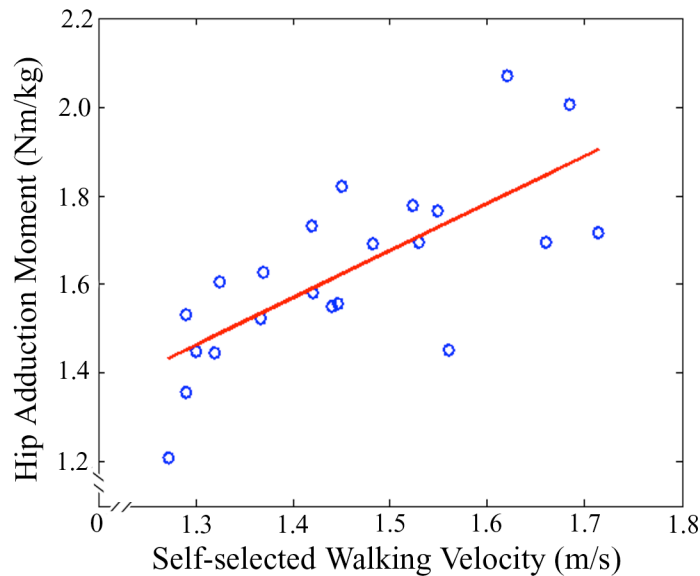
Figure 3: The association between self-selected walking velocity and the peak net external hip adduction moment normalized to body mass captured during the stance phase of the gait cycle r= 0.724, *P*=0.000.

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