EVIDENCE OF ISOMETRIC FUNCTION OF THE FLEXOR HALLUCIS LONGUS AND FLEXOR DIGITORUM LONGUS DURING THE STANCE PHASE OF GAIT

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by
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ABSTRACT

The total force acting upon the forefoot during the push off phase of gait is approximately 120% of body weight, with the first and second rays of the foot being the most heavily loaded. *Flexor hallucis longus* (FHL) and *flexor digitorum longus* (FDL) are multi-articular extrinsic foot muscles directly responsible for flexion of the rays of the foot, and have been shown to play a significant role in supporting the longitudinal arch and balancing forefoot loading. These muscles are anatomic synergists and are active at similar times during the gait cycle, but exhibit dissimilar architectural properties such as mass, muscle fiber length, and physiologic cross sectional area. Despite their importance in normal locomotion, there is little direct experimental evidence examining the function of these muscles *in vivo* or *in situ*. To further investigate the function of the FHL and FDL during ambulation, eight cadaver lower extremities were loaded in the RDAS, a previously validated robotic simulator that re-creates the kinematics and muscle activity of the lower extremity during the stance phase of gait. Simulations were performed with the toe flexor muscles under two distinct control strategies: force-feedback control (FC), where dynamic tendon forces were matched to instantaneous target forces derived from electromyography (EMG) of normal subjects; and isometric displacement control (IDC), where the myotendinous junction of the tendon was held in constant position over the entire stance phase of gait. Distributions of plantar pressure were measured during the simulations to assess forefoot loading. It was hypothesized that instantaneous FHL tendon forces under IDC would be similar to those estimated from EMG data, but FDL forces would not, based on architectural differences between the two muscles. It was also hypothesized that simultaneous IDC (SIDC) of the toe flexor muscles would result in significantly higher tendon forces and forefoot loading compared to FC. Excursions of the FHL and FDL under FC were small (7.18±1.75, 6.32±1.74 mm) relative to optimal muscle fiber length (52.7 mm, 44.6 mm). Instantaneous tendon forces in both the FHL and FDL under IDC were not remarkably different than target forces derived from EMG data. In addition, there were no differences in plantar pressure distributions between FC, IDC or SIDC conditions. This study provides *in situ* experimental evidence of normal forefoot loading under isometric control of the extrinsic toe flexor muscles during stance, suggesting that the FHL and FDL function isometrically during stance phase *in vivo*, an efficient strategy that minimizes energy expenditure and reduces the complexity of their neural control. These results are consistent with the long-standing hypothesis that multi-articular muscles function isometrically to redistribute segmental energy in a fashion that may not possible with mono-articular muscles. The FHL and FDL might be controlled *in vivo* via a local length-servo mechanism such as the stretch reflex; whereby conditions such as peripheral neuropathies (*e.g.*, spastic paresis), or other diseases in which the stretch reflex is impaired (*e.g.*, diabetes) may disrupt normal function. Similarly, any surgical intervention that affects myotendinous length or displacement, or alters the anatomy or geometry of the FHL and FDL might compromise efficient locomotion.
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>List of Figures</td>
<td>vi</td>
</tr>
<tr>
<td>List of Tables</td>
<td>vii</td>
</tr>
<tr>
<td>Acknowledgements</td>
<td>viii</td>
</tr>
<tr>
<td>Chapter 1 Introduction</td>
<td>1</td>
</tr>
<tr>
<td>1.1 Background</td>
<td>1</td>
</tr>
<tr>
<td>1.2 Objectives</td>
<td>4</td>
</tr>
<tr>
<td>1.3 Hypotheses</td>
<td>5</td>
</tr>
<tr>
<td>Chapter 2 Literature Review</td>
<td>6</td>
</tr>
<tr>
<td>2.1 Anatomy</td>
<td>6</td>
</tr>
<tr>
<td>2.2 Function of Multi-articular Muscles</td>
<td>10</td>
</tr>
<tr>
<td>2.3 Toe Flexor Mechanics</td>
<td>14</td>
</tr>
<tr>
<td>2.4 Forefoot Joint Mechanics</td>
<td>18</td>
</tr>
<tr>
<td>2.4.1 Tarsometatarsal Joints</td>
<td>19</td>
</tr>
<tr>
<td>2.4.2 Metatarsophalangeal Joints</td>
<td>21</td>
</tr>
<tr>
<td>2.5 Pathologies of the Forefoot</td>
<td>23</td>
</tr>
<tr>
<td>2.5.1 Flexor tendon pathologies</td>
<td>23</td>
</tr>
<tr>
<td>2.5.2 Hallux Rigidus</td>
<td>24</td>
</tr>
<tr>
<td>2.5.3 Other Disorders</td>
<td>25</td>
</tr>
<tr>
<td>2.6 Summary</td>
<td>26</td>
</tr>
<tr>
<td>Chapter 3 Methods</td>
<td>27</td>
</tr>
<tr>
<td>3.1 Experimental Overview</td>
<td>27</td>
</tr>
<tr>
<td>3.2 Robotic Dynamic Activity Simulator</td>
<td>27</td>
</tr>
<tr>
<td>3.2.1 Kinematic Simulation</td>
<td>28</td>
</tr>
<tr>
<td>3.2.2 Muscle Simulation</td>
<td>29</td>
</tr>
<tr>
<td>3.2.2 Data Acquisition</td>
<td>30</td>
</tr>
<tr>
<td>3.3 Specimen Preparation</td>
<td>30</td>
</tr>
<tr>
<td>3.4 Experimental Protocol</td>
<td>31</td>
</tr>
<tr>
<td>3.4.1 Preliminary Validation Trials</td>
<td>31</td>
</tr>
<tr>
<td>3.4.1 Force-feedback Control Trials</td>
<td>31</td>
</tr>
<tr>
<td>3.4.2 Isometric Displacement Control Trials</td>
<td>32</td>
</tr>
<tr>
<td>3.4.3 Plantar Pressures</td>
<td>32</td>
</tr>
<tr>
<td>3.5 Data Processing</td>
<td>34</td>
</tr>
<tr>
<td>3.6 Statistical Analyses</td>
<td>35</td>
</tr>
<tr>
<td>Chapter 4 Results</td>
<td>36</td>
</tr>
<tr>
<td>4.1 Tendon Excursions</td>
<td>36</td>
</tr>
<tr>
<td>4.1.1 Force-feedback Control</td>
<td>36</td>
</tr>
</tbody>
</table>
4.1.2 Isometric Displacement Control..........................................................40
4.2 Tendon Forces..........................................................................................42
  4.2.1 Force-feedback Control.................................................................42
  4.2.2 Isometric Displacement Control.......................................................44
4.3 Plantar Pressures ..................................................................................49
  4.3.1 Statistical Analyses .........................................................................52

Chapter 5 Discussion ................................................................................54
  5.1 Hypotheses..........................................................................................54
  5.2 Implications..........................................................................................58
  5.3 Limitations of Study Design.................................................................62
  5.4 Future Work .......................................................................................63
  5.5 Concluding Remarks .........................................................................64

References ..................................................................................................65

Appendix A  EMed Software Manuals .......................................................71
LIST OF FIGURES

Figure 1-1. Posterior compartment of the shank, illustrating the origins and tendon paths of the FHL and FDL [Drake et al. 2010]. .................................................................3

Figure 2-1. Dorsal surface (Superior view) of the foot [Drake et al. 2010]. ......................7

Figure 2-2. Plantar view of the foot illustrating the insertions of the FHL and FDL tendons [Drake et al. 2010]. ....................................................................................9

Figure 2-3. EMG activity over one complete gait cycle from initial contact (IC) throughout Stance (St) and Swing (Sw) phases. EMG is expressed as a percentage of maximum muscle test [Perry 1992] ...........................................................................15

Figure 3-1. Experimental setup of the RDAS specimen carriage. 1 - Linear actuators that simulate kinematics, 2 - Load cells and displacement transducers, 3 - Force plate and pressure mat, 4 - Linear actuators that simulate muscle forces, 5 - Liquid nitrogen tubes. .................................................................................................................................28

Figure 3-2. Planar pressure distribution of Specimen #2 under force feedback control, with a mask applied illustrating the regions of interest of the forefoot. M03 - MTH1, M04 - MTH2, M05 - Toe1, M06 - Toe2, M07 - Toe345, M08 - MTH345. ..................34

Figure 4-1. FHL and FDL excursion range for each specimen under force feedback control, averaged over three trials. ...................................................................................36

Figure 4-2. FHL excursion range for each specimen under force feedback control, averaged over three trials. Note: positive excursion values are indicative of a distal displacement, representing simulated muscle lengthening .............................................37

Figure 4-3. FDL Tendon excursion for each specimen under force feedback control, averaged over three trials. ..........................................................................................38

Figure 4-4. FHL excursion vs. FDL excursion, with line of best fit, for eight specimens under force feedback control (r = 0.756, R² = 57.17%). .........................................................38

Figure 4-5. Average FHL tendon excursion (±95% confidence interval) under force feedback control (n=8). ........................................................................................................39

Figure 4-6. Average FDL tendon excursion (±95% confidence interval) under force feedback control (n=8). ..............................................................................................40

Figure 4-7. Comparison of mean tendon excursion (± 1 SD) for FHL and FDL across experimental conditions (n=8). Star denotes a significant (p<0.001) difference of IDC (isometric displacement control) or SIDC (simultaneous IDC) with respect to FC (force feedback control) ..................................................41

Figure 4-8. Tendon excursions (±95% confidence interval) under isometric displacement control (individual and simultaneous) tendon excursions ..............................................42
Figure 4-9. Actual and desired FHL tendon force (±95% confidence interval) under force feedback control.................................................................43

Figure 4-10. Actual and desired FDL tendon force (±95% confidence interval) under force feedback control.........................................................................44

Figure 4-11. FHL and FDL tendon forces under individual isometric displacement control compared to target forces.........................................................45

Figure 4-12. FHL and FDL tendon forces under simultaneous isometric displacement control compared to target forces................................................................46

Figure 4-13. Regression analyses of actual force versus target force for the FHL under FC, IDC, and SIDC for Specimen #2.................................................................47

Figure 4-14. Regression analyses of actual force versus target force for the FDL under FC, IDC, and SIDC for Specimen #2........................................................................48

Figure 4-15. Contact areas (±1 SE) for each region of the foot across experimental conditions.............................................................................................................49

Figure 4-16. Maximum force (±1 SE) for each region of the foot across experimental conditions........................................................................................................50

Figure 4-17. Peak pressures (±1 SE) for each region of the foot across experimental conditions........................................................................................................50

Figure 4-18. Pressure time integrals (±1 SE) for all mask regions across experimental conditions........................................................................................................51

Figure 4-19. Force-time integrals (±1 SE) for all mask regions across experimental conditions........................................................................................................52
LIST OF TABLES

Table 3-1. Information about the eight specimens used. Simulated body weight (BW) was selected based on the robustness of the specimen, and all GRF profiles and force measured to scaled by this value. ................................................................. 31

Table 4-1. Results of regressing FHL target force on actual force for each experimental condition. $B_0$ is the y-intercept of the regression line, $\beta_1$ is the slope, SE is the standard error of the residuals, and $R^2$ is the coefficient of determination. ‘ALL’ represents the representative mean force-tendon curve. ................................................. 47

Table 4-2. Results of regressing FDL target force on actual force for each experimental condition. $B_0$ is the y-intercept of the regression line, $\beta_1$ is the slope, SE is the standard error of the residuals, and $R^2$ is the coefficient of determination. ‘ALL’ represents the representative mean force-tendon curve. ................................................. 48

Table 4-3. General linear model ANOVA p-values for each response variable and each region of the foot ............................................................................................................... 52

Table 4-4. Statistical powers for each response variable and each region of the foot. .......... 53
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Chapter 1

Introduction

1.1 Background

The foot serves as the interface between the lower limbs of the body and the ground during locomotion. The total force acting upon the forefoot during the push off phase of gait is approximately 120% of body weight (Stokes et al. 1979, Hetherington et al. 1992, Hyafune et al. 1999) with the first and second rays of the foot being the most heavily loaded during normal human locomotion (Jacob 2001). Flexor hallucis longus (FHL) and flexor digitorum longus (FDL) are extrinsic foot muscles directly responsible for flexion of the rays of the foot. These muscles have been shown to play a significant role in supporting the longitudinal arch (Thordarson et al. 1995) and optimizing forefoot loading (Kirane et al. 2008b, Ferris et al. 1995). Other passive and active components of toe flexion (the plantar fascia and intrinsic toe flexor muscles, respectively) have also been shown to have a significant effect on the distribution of loads across the forefoot at terminal stance (Hamel et al. 2001). Clearly, the anatomy of the foot has evolved in such a way that allows for an efficient distribution of ground reaction forces during walking (Ridola et al. 2007); however, the mechanics of the FHL and FDL during human locomotion in vivo are not well understood. A more comprehensive understanding of the function and control strategies of extrinsic muscles during human locomotion can play an important role in injury prevention and the detection of foot pathologies.

The FDL originates on the medial side of the posterior surface of the tibia and inserts on the plantar surface of distal phalanges of the lateral four toes (Figure 1-1). In addition to flexion of the lateral four toes, FDL also partially contributes to ankle plantarflexion and subtalar
inversion. The FHL originates on the posterior surface of the fibula and the adjacent interosseous membrane and inserts on the plantar surface of the distal phalanx of the great toe. The primary function of the FHL is to flex the great toe, but it also contributes to ankle plantarflexion and subtalar inversion (Drake et al. 2010). Thus, the FHL and FDL are anatomic synergists, i.e., they are muscles that assist each other in performing a specific joint action or collection of joint actions.

The architectural properties of muscle are generally assumed to be a strong predictor of how the muscle functions, with muscle fiber length considered to be one of the most powerful predictors (Lieber et al. 2011). The FHL has a greater fiber length (52.7 mm vs. 44.6 mm) mass (38.9 g vs. 20.3 g) and higher physiological cross-sectional area (PCSA; 7.2 cm² vs. 4.5 cm²) compared to the FDL (Ward et al. 2009). This suggests that they may function differently during normal locomotion, despite the fact that both are mutually innervated by the tibial nerve and electromyography (EMG) readings have found that these two muscles are active at similar times during the gait cycle (Perry 1992; Mann & Hagy 1983; Sutherland 2001). Both muscles have a very large distal tendon slack length (FHL - 378 mm, FDL - 356 mm) relative to muscle fiber length (Arnold et al. 2010).

Many investigations have examined the activity of FHL and FDL in feline hindlimbs. Despite being near mechanical synergists and mutually innervated, FHL and FDL were found to exhibit significantly different EMG patterns during quadruped locomotion (O’Donovan et al. 1982). Fictive locomotion (i.e., stimulation of the locomotor region in the mesencephalon of decerebrate animals) was elicited in cats with similar results, leading the researchers to conclude that there were differentiated neural drives to the FHL and FDL (Fleshman et al. 1984; Burke et al. 2001). Although it is assumed that there are no significant differences in the neural networks between man and other vertebrates, there are differences in the motor output responsible for locomotion across species (Duysens & Van de Crommert 1998).
Figure 1-1. Posterior compartment of the shank, illustrating the origins and tendon paths of the FHL and FDL. [Drake et al. 2010].
It has been hypothesized that multi-articular muscles function isometrically during certain tasks, perhaps to redistribute and exchange energy amongst body segments in ways that may be impossible by mono-articular muscles. This is because mono-articular muscles can only induce a joint rotation in the same direction of the only joint moment that it causes, unlike a multi-articular muscle that directly contributes to multiple joint moments (Zajac et al. 2002). Despite the fact that these muscles play a critical role in locomotion, there is little direct experimental evidence examining their function in vivo or in situ. A previous experiment in our lab provided evidence that the multi-articular FHL functions isometrically during cadaveric simulations of the stance phase of gait (Kirane et al. 2008). This study did not investigate the mechanics of the FDL, and the power of the statistical results was limited by a small sample size.

1.2 Objectives

The goals of the present investigation are to examine the mechanics of the multi-articular extrinsic toe flexors during stance by: 1. validating previous results related to the potential isometric functioning of the FHL during walking, 2. investigating the function of FDL during stance phase of gait, and 3. proposing potential control strategies consistent with our results. This study will utilize the Robotic Dynamic Activity Simulator (RDAS) to simulate the kinematics and muscle activity of lower extremity cadaveric specimens during the stance phase of gait (Sharkey & Hamel 1998, Hoskins 2006). The RDAS simulates muscle function via linear actuators fixed to the tendons of the extrinsic muscles of the shank at their myotendinous junctions, and it has the capability to do so in two unique control strategies. The first is termed force-feedback control (FC), where the instantaneous muscle forces are matched to target forces derived from EMG patterns of live subjects. The second is isometric displacement control (IDC), where the controller seeks to maintain a stationary myotendinous junction (i.e., hold the tendon at a constant
position. The main objectives of this experiment are to: 1. dynamically measure the forces in the FHL and FDL tendons, and 2. assess forefoot loading using plantar pressure distributions to examine if there is any change in forefoot mechanics under these two different muscle control strategies.

1.3 Hypotheses

_Hypothesis 1:_

The FHL functions isometrically during the stance phase of gait, with the geometry and kinematics of the foot dictating functionally appropriate tendon force.

_Hypothesis 2:_

The FDL does not function isometrically during the stance phase of gait; using this control strategy will produce functionally inappropriate tendon forces.

_Hypothesis 3:_

Simultaneous isometric control of the FDL and FHL during the stance phase of gait will result in a significant increase in tendon forces and forefoot loading.
Chapter 2

Literature Review

2.1 Anatomy

The human foot is the most distal segment of the lower extremity and is comprised of 26 bones with an additional two sesamoid bones. The primary functions of the foot are weight bearing and locomotion. The foot can be divided into three functional segments: the forefoot, midfoot, and hindfoot. Starting distally at the toes, the great toe consists of two bones, the proximal and distal phalanges. The remaining four rays are comprised of three bones, the proximal, intermediate, and distal phalanges. Within the lateral four rays, the articulations of the proximal and intermediate phalanges are called the proximal interphalangeal (PIP) joints, while the joints between the intermediate and distal phalanges are referred to as distal interphalangeal (DIP) joints. Each of five metatarsals (I-V) articulates with the proximal phalange of each ray at the metatarsophalangeal (MTP) joints. This group of five digits and metatarsal bones collectively form the forefoot (Figure 2-1).

The bones of the midfoot form arches that aid in the weight bearing capacity of the foot. Cuneiforms I, II, and III articulate with the proximal ends of metatarsals I, II, and III, while the proximal end of metatarsals IV and V articulate with the cuboid. These articulations are referred to as the tarsometatarsal (TMT) joints. Medial to the cuboid and proximal to the cuneiforms is the navicular, or intermediate tarsal bone.

The talus and calcaneus comprise the hindfoot and lie anterior and posterior to the navicular, respectively. The talus articulates with the distal tibia to form the talocrural joint and the distal tibia superiorly to form the talocrural joint and calcaneus inferiorly to form the subtalar
joint. These three bones form the articulations about which ankle movements are performed. Dorsiflexion and plantarflexion occur at the talocrural joint while inversion and eversion of the hindfoot occur at the subtalar joint.

Figure 2-1. Dorsal surface (Superior view) of the foot [Drake et al. 2010].

The bones of the foot are held together rigidly with ligaments and form two arches relative to the ground when standing upright. The transverse arch runs medial to lateral and spans from the talus posteriorly, to the metatarsal heads, anteriorly. The longitudinal arch runs anterior to posterior, and is formed from the plantar surface of the calcaneus to the metatarsal heads. It can be divided into medial and lateral portions, with the medial arch being more pronounced. These arches can be approximated by a curved beam. The longitudinal arch has been functionally described as a ‘windlass mechanism’, where the bones of the midfoot represent members of a
truss and the plantar fascia represents the tie-rod. The main function of the arches is to stabilize the foot while distributing loads due to body weight during standing and locomotion, which is explained by this mechanical representation.

Two classes of muscles contribute to motions critical for human locomotion. Intrinsic muscles originate and insert exclusively within the foot, while extrinsic muscles originate more proximally on the lower extremity and insert within the foot. The intrinsic muscles are responsible for fine-tuning the actions of the long tendons of the extrinsic muscles. Extrinsic flexor muscles, specifically flexor hallucis longus (FHL) and flexor digitorum longus (FDL), are the primary muscles responsible for flexion of the great toe and remaining four digits.

A collection of connective tissues, the flexor retinaculum, extends from the medial malleolus to the medial surface of the calcaneus; it allows for the routing of the toe flexor tendons from the posterior compartment of the leg, around the talocrural and subtalar joints, to the plantar surface of the foot. The flexor retinaculum is a continuation of the deep fascia of the anterior leg and the plantar aponeurosis, a thick and fibrous layering of deep fascia of the sole of the foot. The plantar fascia, of which the plantar aponeurosis is a central portion, supports the longitudinal arch. It originates from the medial process of the calcaneal tuberosity and divides into five segments at approximately the midpoint of the sole that continue anteriorly in the direction of the five rays of the foot. These segments are superficially crossed by retinacular bands forming the transverse metatarsal ligaments.

At the distal end of the tibia, the FHL tendon is separated from the FDL by an intertendinous space. Distal to the talocalcaneal joint and at the end of the calcaneal tunnel, the tendons of the FDL and FHL merge to a common sheath. This location is referred to as the master knot of Henry. Following this convergence, FDL tendon crosses plantar to FHL tendon as they continue anteriorly to their respective insertions. Anterior to this crossing, the FHL anastomoses (i.e., connects via a tendinous slip) with the FDL. This connection typically occurs
at the location of the divergence of the FDL into four tendons, which are numbered corresponding to the four rays (II-V) on which they insert (Figure 2-2). Although some anatomic variability exists in the anastomoses (LaRue & Anctil 2006), in most cases the tendon slip connects the FHL tendon to the second and third tendons of the FDL.

Figure 2-2. Plantar view of the foot illustrating the insertions of the FHL and FDL tendons [Drake et al. 2010].

Several intrinsic foot muscles interact with the tendons of FDL and FHL. The quadratus plantae originates on the calcaneal tuberosity and inserts directly on the lateral side of the FDL tendon, anterior to its four segment division, allowing it to assist in the flexion of the lateral four rays. In addition, each of FDL’s four associated tendons give origin to the lumbrical muscles, which insert on the extensor hoods of rays II-V and assists in flexion of the corresponding MTP joints. Similar to the FDL, the flexor digitorum brevis flexes the lateral four toes, although its
insertion is on the intermediate phalanges; the *flexor hallucis brevis* inserts on the proximal phalange of the great toe and also aids in flexion at the MTP joint. The majority of the remainder of intrinsic foot muscles play a role in abduction, adduction, or extension of the toes (Drake *et al.* 2010, Sarrafian 1993).

Flexion of the toes is a critical component of human locomotion. Although the primary toe flexors originate in the posterior compartment of the leg, numerous intrinsic muscles also contribute to toe flexion. The structure and function of the foot has evolved to distribute the high loads experienced during normal standing and locomotion in an optimized fashion.

### 2.2 Function of Multi-articular Muscles

Skeletal muscles are commonly classified by the arrangement and orientation of their fibers, anatomic location of origin and/or insertion, fiber type (*i.e.*, fast or slow twitch), and by the movements to which they contribute. Multi-articular muscles span more than one joint, simultaneously contributing to multiple joint torques. The foot includes a number of multi-articular extrinsic muscles that contribute to joint movements necessary for locomotion. Multi-articular FHL and FDL originate in the posterior compartment of the leg and span the joints of the ankle, the relatively immobile joints of the midfoot, the MTP, PIP and DIP joints before inserting on the plantar surface of the distal phalanges. A better appreciation of how these multi-articular muscles function during gait may provide insight into motor control and movement coordination, and the pathologies affecting them.

van Ingen Schenau *et al.* (1987) recorded kinematics and muscle activity of the lower extremity during a vertical jump in an attempt to explain a potential functional role of bi-articular muscles. During this type of motion, knee angular velocity decreases as the knee approaches full extension, a phenomenon that serves as a protection to maintain joint continuity and prevent knee
hyperextension. It was found that the bi-articular *gastrocnemius* increased in EMG activity at the peak knee extension angular velocity, thereby producing a rigid link to transfer additional positive work from knee extensors (*rectus femoris, vastus lateralis*) to the ankle for plantar flexion during the push-off phase of the vertical jump. Although the angular velocity approaches zero as extension proceeds, the knee extensors were still able to deliver power for plantar flexion due to the increase in activation of the *gastrocnemius*. This study provided evidence that the action of bi-articular muscles may contribute to a proximal-to-distal transfer of power.

van Ingen Schenau *et al.* (1992) expanded upon investigations of multi-articular muscles by conducting kinematic and kinetic analyses of cycling. It was hypothesized that mono- and bi-articular muscles play unique roles in the control of movement coordination. Inverse dynamics were performed based on force measurements from transducers within the pedals and lower extremity EMG measurements. Muscle forces, joint moments and powers, as well as muscle shortening velocities were estimated. For mono-articular muscles, periods of increased mechanical response and increased shortening velocity occurred almost exclusively during hip, knee, and ankle plantarflexion. This suggested that mono-articular muscles were active during periods of time where they mainly shortened, thus contributing to positive work output. Bi-articular muscles were found to be active mainly during shifts in joint moments corresponding to changes in the direction of force exerted on the crank. This is consistent with the idea that the directional control of external forces during multi-joint movements requires a particular set of joint moments; bi-articular muscles appeared to function to alter this net joint moment distribution, thereby prescribing the external force direction. These results agreed with that of other studies (Jacobs *et al.* 1993, van Bolhuis *et al.* 1998) and suggested that there may be unique fundamental differences between the functioning of mono- and bi-articular muscles.

Prilutsky & Zatsiorsky (1994) recorded EMG, kinematic, and kinetic data during jumping, landing, and jogging in an effort to describe the potential isometric functionality of
lower extremity bi-articular muscles and the roles they play in the transfer of mechanical energy. A mathematical musculoskeletal model was derived and utilized to calculate joint moments, and subsequently, muscle forces based on joint locations and ground reaction force input. Energy transfer was then computed by taking the integral of the power developed by the joint moments and forces. The results suggested that there was a net transfer of mechanical energy proximal to distal during jumping and running. It was suggested that this may be due to physiologic differences between proximal and distal muscles. Proximal muscles typically have greater volume and cross sectional area, making them more suitable for force production. When functioning isometrically, also described as ‘tendinous action’ (van Ingen Schenau et al. 1990), bi-articular muscles may allow proximal muscles to compensate for smaller distal muscles by transferring energy amongst joints. In a similar study, Jacobs et al. (1996) examined one-legged jumping and sprinting in an attempt to quantify the contributions of six muscle groups (bi-articular rectus femoris, gastrocnemius, and hamstrings; mono-articular soleus, vasti, and gluteus maximus) to the transfer of power between joints of the lower extremity. A Hill-based muscle model was utilized to predict muscle forces and it was concluded that a net transfer of power was delivered from the hip to the knee due to the action of bi-articular muscles.

van Bolhuis et al. (1998) attempted to investigate the hypothesis that mono- and bi-articular muscles play specialized roles in controlling the direction and magnitude of a desired externally applied force. The force and movement direction of the upper arm were varied independently, and EMG was recorded from muscles of the upper extremity while the arm moved in a horizontal plane. Results showed that bi-articular muscle activation was based solely on direction of the externally applied force vector, whereas mono-articular muscle activation correlated with both the direction of the movement and direction of the force.

In a review, Zajac et al. (2002) caution against making generalizations regarding movement coordination based on functional differences between mono- and multi-articular
muscles. The contraction of muscles produces forces that cause linear accelerations of body segments and angular accelerations of joints. Due to dynamic coupling, these motions need not be limited to the segments to which the muscles originate or insert (Zajac 1993). This is because all muscle forces acting on a kinematic chain contribute to all joint intersegmental forces; both uni-articular and bi-articular muscles can play roles in the transfer of power (Neptune et al. 2001, Neptune et al. 2004). Bi-articular muscles might, however, allow for segmental energy flow in a unique fashion not possible by mono-articular muscles. Additionally, multi-articular muscles may function by accelerating a body segment in directions other than the principal force producing direction of an associated mono-articular muscle for that segment. This may be due to the fact that resultant forces generated by bi-articular muscles can have a significant component orthogonal to the body segments, whereas mono-articular muscles’ line of action is typically in the along the axis of the segment (Hof 2001). Clearly, the role of multi-articular muscles in multi-joint movements still requires investigation. It has been suggested that computer-based dynamic simulations of musculoskeletal models can provide results necessary to further elucidate the roles that multi-articular muscles play in the coordination of motor tasks such as locomotion (Neptune et al. 2009).

The human body can perform a seemingly infinite number of complex, highly coordinated movements. Activation of muscles by the central nervous system (CNS) produces the mechanical output needed to move the body’s segments; however, it is still not fully understood what methods or strategies the CNS routinely employs in the control of movement. Multi-articular muscles may provide the CNS an opportunity to simplify the signal processing necessary for certain motor functions. Evolutionarily speaking, the existence of bi-articular muscles suggests that they should provide a functional advantage (e.g., decreased energetic cost, increased complexity of mechanical output) which an entire body of mono-articular muscles cannot provide (Bobbert & van Soest 2000). In the case of the multi-articular extrinsic toe
flexors, their origins within the posterior shank enable controlled and powerful flexion of the toes without adding tremendous muscle mass to the foot. Although many studies have illustrated functional differences between mono- and multi-articular muscles, additional research is necessary to further elucidate biomechanical, metabolic, or neural control advantages offered by multi-articular muscles.

2.3 Toe Flexor Mechanics

Proper functioning of the FDL and FHL is critical to normal locomotion. Human walking is functionally broken down into phases, collectively referred to as the gait cycle (GC). A particular limb goes through two main phases: stance phase and swing phase, comprising of approximately 60% and 40% of the GC, respectively. Herein, a single GC will be assumed to begin at the instant when the heel makes contact with the ground (heel strike), and percentages will be representative of the proportion of total time of the GC, starting with this initial contact. Stance phase begins with heel strike, and is the period of the GC when a particular limb is in contact with the ground. Stance phase ends at toe-off, the instant where the rays of the foot leave the ground. This signifies the beginning of swing phase, the period in which the limb is forwardly progressing when not in contact with the ground. Swing phase ends at the next heel strike, and the cycle is completed. Each limb progresses through this same cycle, but the phases for opposing limbs are temporally shifted. The time duration of stance and swing phases are not equal, as a result, there are periods of time during stance phase where the other limb is also in contact with the ground, referred to as double support. However, the majority of stance phase is spent in single support, where one limb is supporting the weight of the body. The previous phases can be further subdivided into pre-, initial, mid-, and terminal phases. The roles of the lower extremities in contact with the ground during the gait cycle are to propel the body forward,
maintain postural stability, minimize impact transients from heel strike, and minimize energy expenditure (Perry 1992).

EMG readings from indwelling electrodes indicate that FDL is typically active from mid-stance (~10% of GC) to the initiation of pre-swing (~52% of GC). FHL is typically active from the latter stages of mid-stance (~25% of GC) to the initiation of pre-swing (~52% of GC). Both FDL and FHL reach their relative peak EMG toward the end of terminal stance, just prior to toe-off. The FDL peak is approximately 40% of its maximal voluntary contraction (MVC) while FHL peaks at around 80% MVC (Perry 1992, Figure 2-3).

Figure 2-3. EMG activity over one complete gait cycle from initial contact (IC) throughout Stance (St) and Swing (Sw) phases. EMG is expressed as a percentage of maximum muscle test [Perry 1992].

Jacob (2001) utilized anthropometric data and forces collected underneath the forefoot by means of a pressure mat to estimate forefoot mechanics during push-off. FHL was found to have an approximate tendon force of 52.4% of body weight (BW), while the tendon force of the FDL along the second ray was approximated as 8.8% BW. Additional foot muscle forces calculated include: flexor hallucis brevis (35.5% BW), peroneus longus (57.8% BW), and flexor digitorum brevis (12.5% BW). Clearly evident, many muscles of the foot (especially toe flexors) are subject to high loads during push-off, a critical phase in the gait cycle for forward propulsion. In
addition, these data show that the intrinsic *flexor brevis* muscles might have a significant contribution to balancing forefoot loading, as is the case with the *flexor longus* muscles (Ferris *et al.* 1995).

Hintermann *et al.* (1994) measured tendon excursions with respect to eversion-inversion (EI) and plantarflexion-dorsiflexion (PD) axes of the foot with an apparatus that secured lower extremity cadavers while allowing physiologic ranges of motion. The specimens were moved through a range of motion of -9° eversion to 21° inversion, and -20° dorsiflexion to 30° plantarflexion, all with respect to the neutral anatomic position of the foot. FHL was found to exhibit excursion ranges of 10.1 mm and 16.9 mm with respect to EI and PD axes of the foot, respectively. FDL exhibited 12.2 mm and 12.4 mm tendon excursion with respect to EI and PD axes of the foot, respectively. No toe movement was possible with this experimental set up, which would understandably have a significant effect on the overall excursions of FDL and FHL.

Kirane *et al.* (2008) dynamically recorded tendon excursions in cadavers during simulated gait trials utilizing the Robotic Dynamic Activity Simulator (RDAS). Across five specimens, average tendon excursions for FHL were reported as 6.57 ± 3.13 mm while an average of 6.62 mm was reported for FDL. These excursions are smaller than previously reported, as full ranges of eversion-inversion and dorsiflexion-plantarflexion are not common during stance phase of gait. Flexion and extension about the MTP joints would certainly affect the excursions of FHL and FDL, and provides a better representation of *in vivo* behavior.

The FDL and FHL are primarily toe flexors, but they also contribute to plantarflexion and subtalar inversion. A number of research groups (*e.g.*, Hintermann *et al.* 1994, Spoor *et al.* 1990, Fick 1911, Ambagtsheer 1978) have attempted to estimate moment arms of the FDL and FHL with respect to the primary axes of motion within the foot-ankle complex (*viz.* evertor-invertor axis and dorsiflexion-plantarflexion axis). A review compiled by Hintermann *et al.* reported values of evertor-invertor moment arm (MA_{EI}) as a fraction of the movement produced by *tibialis*
posterior. The dorsiflexion-plantarflexion moment arm (MA_{DP}) were given as ranges starting with the value at full dorsiflexion (-20°) and ending with the value at full plantarflexion (30°). Fick (1911, from: Hintermann et al. 1994) recorded MA_{EI} for the FDL as 0.90 and FHL as 0.57. Ambagtsheer (1978, from: Hintermann et al. 1994) recorded the MA_{EI} of the FDL as 0.71 and the FHL moment arm as 0.29. Finally, Hintermann et al. (1994) reported an MA_{EI} of 0.75 and 0.62 for the FDL and FHL, respectively. Varying definitions of ‘evertor-invertor axis’ most likely explain the disparity amongst reported values. For example, Fick recorded moment arm values with respect to the subtalar joint axis. Eversion and inversion are complex movements that do not occur solely about the subtalar joint axis, but about multiple joint axes of the foot-ankle complex. Hintermann measured MA_{EI} about an instantaneous axis of motion, and therefore, the data are probably more reliable than other groups.

Spoor et al. (1990) estimated the range of MA_{DP} for FDL and FHL as 23mm-13mm and 37mm-19mm, respectively. Conversely, Hintermann et al. (1994) found ranges of MA_{DP} of 10mm-19mm and 15mm-24mm for the FDL and FHL, respectively. The disparity of the above results can be accounted for by differences in experimental methodology amongst the research groups. Spoor et al. mounted specimens in such a fashion that did not allow sagittal plane movements and restricted tibial rotation and foot eversion-inversion. This most likely explains the differing trends amongst MA_{DP} when progressing throughout the full range of motion between research groups.

Reeser et al. (1983) recorded EMG from intrinsic and extrinsic toe flexors by use of indwelling electrodes during a variety of movement tasks including: walking, stepping, jogging, and running. Specifically, the study examined the roles that quadratus plantae (QP) and flexor digitorum brevis (FDB) play in assisting the FDL in toe flexion during times of ‘active insufficiency’ i.e., foot positions in which the optimal functioning of the FDL is compromised such as inversion or plantarflexion. In both situations, the FDB and QP significantly increased in
activity, leading to the conclusion that there is interactive behavior amongst these toe flexors. The research group also found that the toe flexors increase in activity nearing the push-off phase of gait in an effort to oppose the inevitable MTP joint extension that would occur as the ground reaction force is transferred to the forefoot. This increase in activity by the intrinsic toe flexors also serves to provide additional flexing force when required by a particular activity.

FDL and FHL exhibit peak activity at the final stages of the stance phase of gait, a time in which the forefoot is subject to significant loading. Toe flexion at this point in the GC is important for forward propulsion, and consequently, the distribution of loading across the forefoot. Intrinsic muscles have been shown to have significant interactions with the long toe flexors, and play important roles in optimizing the loading distribution across the forefoot (Reeser et al. 1983).

2.4 Forefoot Joint Mechanics

The human foot is divided into three functional parts: the forefoot, midfoot, and hindfoot. The hindfoot, comprised of the calcaneus and talus, serves as the interface between the lower leg and the foot. The hindfoot together with the lower leg (i.e., distal tibia and fibula) is often referred to as the foot-ankle complex. The bones of the midfoot form the transverse arches of the foot, and transmit loads between the forefoot and hindfoot. It consists of the navicular, the cuboid, and the three cuneiforms. The forefoot begins proximally at the tarsometatarsal (TMT) joints, the articulations of the proximal ends of the metatarsals with the cuneiforms and cuboid. The metatarsophalangeal (MTP) joints are the articulations between the distal ends, or heads, of the metatarsals and the proximal phalanges, at the base of the digits. The metatarsals are critical components of the foot’s longitudinal and transverse arches, and serve to transfer and distribute loads while walking over uneven surfaces (Perry 1992, Drake et al. 2010). The remainder of this
section will focus on the kinematics and kinetics of the TMT and MTP joints of the foot, and their function during normal locomotion.

2.4.1 Tarsometatarsal Joints

The most proximal region of the forefoot consists of the TMT joint articulations. The first and second TMT joints are collectively referred to as the Lisfranc joint complex, which is held together rigidly by several strong ligaments. This region is composed of several intermetatarsal and inter-tarsal joints, although the first and second metatarsals do not typically articulate with one another. There is significant variability in many of the anatomic parameters of the Lisfranc complex; one reason why the ligaments of this region are often described as a ‘ligamentous system’ (de Palma et al. 1997) rather than with discrete nomenclature. This system is divided into three main classifications: dorsal, interosseous, and plantar. A particular interosseous ligament of interest is the Lisfranc ligament, which spans from the medial cuneiform to the second metatarsal. It is the largest and strongest ligament, approximately 9 mm long and 5 mm thick. All articulations of this region are reinforced by ligaments of varying strength, with the overall aim to maintain stability (Solan et al. 2001, de Palma et al. 1997).

Several research groups have investigated the functional range of motion (ROM) of the TMT joints. Ouzounian & Shereff (1989) found a mean of 1.5° supination-pronation and 3.5° of dorsiflexion-planterflexion for the first TMT joint. The second TMT joint exhibited an even lower dorsiflexion-planterflexion ROM of 0.6°, and a similar supination-pronation ROM of 1.2°. The third, fourth, and fifth TMT joints exhibited significantly higher ROM than those of the Lisfranc joint complex. Wanivenhaus & Pretterkleiber (1989) observed a mean value of 4.3° of dorsiflexion, 4.4° of abduction, and 5° of adduction at the first TMT joint. Stiffness in this region,
as characterized by the low ROM of the Lisfranc joint complex, is critical to the transfer of forces between the forefoot and hindfoot during the push-off phases of the gait cycle.

Fauth et al. (2004) attempted to record first and second TMT joint stiffness of intact joints removed from cadaveric specimens. The joints were subjected to varying peak moment levels and resulting angular displacements were measured. The curves of moment vs. angle exhibited an initial region of low, relatively constant stiffness, followed by a transition point into region of drastically increasing stiffness. The angle at which this transition point occurred was used to approximate the normal functional ROM of the joint. The results of this study agree with previously published data regarding the ROM of the first TMT; however, the results suggest that the functional ROM of the second TMT joint may be higher (between 4-5º) than previously published data (Ouzounian & Shereff 1989).

Kinetic analyses of the TMT joints provide insight into the mechanisms responsible for force transmission between segments of the foot. Lakin et al. (2001) approximated the TMT joint pressures by means of pressure sensitive film inserted within the joints of cadaveric specimens under a variety of foot loading conditions and positions. The results suggest that the TMT joints control loading by two different mechanisms. When subjected to low but increasing loads, a corresponding increase in joint contact area served to maintain a relatively constant joint contact pressure. Conversely, when subjected to higher loads, an apparent transfer of loading occurred amongst the TMT joints. This was evidenced by an increase in first and fourth/fifth TMT joint contact forces. Overall these findings suggest that two distinct mechanisms, inter-joint and intra-joint load sharing, are employed for appropriate, non-damaging force transmission up the kinetic chain. Joint contact forces at the second and third TMT joints were significantly higher when compared to the first and fourth/fifth TMT joints across almost all levels of loading and foot positions. Throughout all trials, the third TMT joint demonstrated the highest joint contact pressures suggesting that it, in addition to the Lisfranc joint complex, plays a significant role in
the control of loading across the foot. It should be noted, however, that these specimens were subject to passive loading with no simulated muscle function. Thus, the observed magnitudes and distributions of TMT joint contact forces may be different than of those observed in vivo.

2.4.2 Metatarsophalangeal Joints

The first MTP joint is a unique joint for several reasons. The range of dorsiflexion is of much greater significance than the range of plantarflexion, as the hallux is in dorsiflexion during most activities. The first MTP joint consists of three distinct articulations, all of which have been shown to have an effect on the loads experienced at the joint, and subsequently, the function of the joint. The complex configuration of the sesamoidal bones inferior to the articulation at the metatarsal head aids in redistributing ground reaction forces at the MTP joint, which are exerted orthogonal to the long axis of the metatarsal (Yoshioka et al. 1988, Shereff et al. 1986). The primary articular surface resides between the metatarsal head and the proximal phalangeal surface, with secondary articulating surfaces between the medial and lateral sesamoidal bones and the grooves of the inferior surface of the metatarsal head (Yoshioka et al. 1988). The instantaneous centers of rotation of all articulations of the first MTP joint (including that of the medial and lateral sesamoids with the inferior metatarsal head) all occur within the metatarsal head, but significant deviations from this trend have been observed in cases of hallux valgus and hallux rigidus (Shereff et al. 1986). Active plantarflexion/dorsiflexion ROM for the first MTP joint has been reported as 26.4°/34.3° (Hetherington et al. 1989) and 21°/51° (Joseph 1954), while passive ROM has been reported as 34°/76° (Shereff et al. 1986) and 45°/70° (Hoppenfeld 1976). The functional ROM with respect to abduction/adduction of the hallux is significantly less, reported as 4.2°/7.2° (Hetherington et al. 1989). The sesamoid bones have been found to
translate along their respective articulating surfaces over an arc of 49º, with a displacement ranging from 10 to 12mm, or about 90% of the length of the sesamoid bones (Shereff et al. 1986).

The motion between the first metatarsal head and proximal phalange articulating surface is described as ‘tangential sliding’ from maximal plantarflexion (about 35º) to approximately 40º of dorsiflexion, while compression between the surface increased when approaching maximal dorsiflexion at approximately 75º. During these motions, deviation in the transverse plane at this articulation was negligible (Shereff et al. 1986). In neutral position, the first MTP joint is in approximately 15-20º of dorsiflexion. Approximately 65º of first MTP extension is necessary during stance phase of the human gait cycle (Hopson et al. 1995).

The force acting on the first metatarsal head has been approximated using a variety of different methods as 120% BW (Jacob 2001, Hetherington et al. 1992, Stokes et al. 1979). The resultant force on the second metatarsal head has been reported as 45.4% BW (Jacob 2001). Stokes et al. (1979) approximated the joint reaction force at the first MTP joint to be 79.9% BW and at the second MTP joint to be 58.3% BW. The first and second metatarsal axial forces were approximated as 134% BW and 84.3% BW, while the first and second metatarsal bending moments were estimated to be 2.0 and 1.4 (%BW*m), respectively. Sharkey et al. (1995) measured axial second metatarsal axial forces to be -450±303 N, while bending moments were measured to be 4.01±3.71 N*m. Jacob (2001) estimated the first and second MTP joint reaction forces as 86.4% BW and 21.5% BW, respectively. Kirane et al. (2008) measured strains across the first metatarsal during simulated gait in cadavers, and subsequently derived first MTP joint reaction forces to be approximately 90% BW. However, it has been suggested (Fauth et al. 2004) that the moments experienced by the first and second metatarsal heads in vivo may be substantially less than the 5 to 13 Newton-meter range that was previously reported by researchers performing mathematical approximations.
The third, fourth, and fifth MTP joint reaction forces have also been approximated from foot geometry. Stokes et al. (1978) approximated the third, fourth, and fifth MTP joint reaction forces as 29.5% BW, 20.9% BW, and 13.9% BW, respectively; Jacobs (2001) reported values of 22.3% BW, 10.3% BW, and 3.5% BW, respectively. Aside from these mathematical approximations, there have been no direct experimental measures of joint reaction forces at the lateral three MTP joints in vivo.

2.5 Pathologies of the Forefoot

2.5.1 Flexor tendon pathologies

FHL tendonitis, inflammation and swelling of the FHL tendon, has been documented to be relatively common among dancers (Hamilton 1977, Hamilton et al. 1988) and athletes (Sammarco 1998). Ballet dancers routinely perform maneuvers that hyper-plantarflex the foot and place high loads across extended rays. This places undue stress on the FHL tendon and can lead to discomfort and injury. Athletes in a number of sports (e.g., soccer, football, basketball) perform bouts of forceful plantarflexion and rapid cutting movements. Similar to dancers, the FHL is subject to high loading scenarios in vulnerable positions and orientations that can result in injury. Overuse has been hypothesized as the primary cause of FHL tendonitis, and can also lead to tears and, although rare, rupture of the tendon (e.g., Krackow et al. 1980). This kind of rupture has been treated in the past by suturing the flexor hallucis brevis tendon to the site of the rupture (Grispigni et al. 2000).

Tenosynovitis, an inflammation of the fluid filled tendinous sheath, has been reported in both the FHL (Gould 1981, Hamilton 1982) and, although very rare, the FDL (Diwanji & Shah 2008). The FHL is subject to restriction at numerous sites including: the fibro-osseous tunnel, the knot of Henry, and at the sesamoids inferior to the first metatarsal head. Restrictions at these
sites can play a role in causing tenosynovitis, and as a result, significantly affect normal range of motion or mechanical functionality. Pain as a result of tenosynovitis of the FHL has been reported as most often occurring in the posteromedial ankle (Michelson & Dunn 2005). Swelling and inflammation restricts tendon excursions and thus, surgical intervention is often recommended when conservative treatment fails.

2.5.2 Hallux Rigidus

Hallux rigidus is one of the most common deformities of the first ray. Hallux rigidus is described as osteoarthritis of the first MTP joint, and is accompanied by a decreased range of dorsiflexion (Bingold & Collins 1950). Although the etiology of hallux rigidus is not clearly defined, researchers have suggested some possible causes \(i.e\.,\) abnormal gait patterns arising to stabilize a hypermobile hallux or injured first MTP joint and increased friction at the articulating surface as a result of a shortened flexor brevis tendon. Studies have also suggested that cartilaginous lesions at the first metatarsal head, arising from injuries including sudden forceful compression, contribute to the pathogenesis of hallux rigidus (McMaster 1978). Michelson & Dunn (2005) described the potential relationship between FHL tenosynovitis and hallux rigidus. In a summary of 81 subjects, the authors concluded that limited FHL excursion may play a role in the development of hallux rigidus. In addition, a condition referred to as ‘pseudo’ hallux rigidus has been characterized as a swelling of the FHL at the fibro-osseus tunnel that results in a loss of hallux dorsiflexion similar to that of true hallux rigidus (Hamilton 1988). Cases of hallux rigidus have been shown to alter normal kinematics and kinetics of the first MTP joint, most likely a result of changes to the articulating surface at the joint (Shereff et al. 1986).
2.5.3 Other Disorders

There are a number of other disorders affecting the forefoot that may be a cause of, or result in, a loss in normal toe flexor function. The sesamoid bones of the MTP joint complex are subject to high loading, and therefore, are susceptible to injuries such as stress fractures and inflammatory conditions. As the sesamoids are embedded within the hallux flexor tendons, sesamoidal pathologies can alter the range of motion of the hallux and have significant effects on locomotor patterns (Cohen 2009).

The second metatarsal is one the most commonly reported sites of stress fracture amongst athletes, and the most commonly reported stress fracture of the foot (Orava 1980, Sullivan et al. 1984). The hallux is subject to high loads during human locomotion, but it is significantly more robust than the lesser metatarsals (Sarrafian 1993). Thus, the lesser metatarsals that experience high forces would be expected to exhibit a greater frequency of stress fractures. Sharkey et al. (1995) found evidence that decreased activity of plantarflexors (including FHL and FDL) may increase bending moments at the second metatarsal and, subsequently, lead to an increase in stress fractures.

Hammer toe and claw toe are two deformities associated with the lesser toes, and are commonly seen in diabetic patients. While both conditions are associated with hyperextension of the MTP joint in neutral position, hammer toe includes abnormal proximal interphalangeal joint flexion (Schrier et al. 2009). It is most commonly accepted that the most significant cause of these deformities is an imbalance of intrinsic and extrinsic muscles spanning the joints of the toe, resulting from injury or motor neuropathy. However, recently published work suggests that muscle atrophy and/or imbalances may not play as significant a role in the etiology of claw toe in the diabetic foot as originally thought (Bus et al. 2009). The researchers recommend that the pathogenesis of such disorders is very complex, and other causes, such as disorders relating to the
plantar aponeurosis or alterations to the MTP joint capsule, might be more significant in the development of these disorders.

2.6 Summary

Analysis of the structure and function of muscles and joints of the forefoot can lend insight to the foot’s remarkable ability to withstand and efficiently distribute the high loads experienced during locomotion. The extrinsic toe flexor muscles are important in maintaining normal locomotion; thus, a better understanding of how these muscles function during gait is needed. Gait simulation using cadaveric specimens have proven valuable and can provide important information that is near impossible to collect from live human subjects.
Chapter 3

Methods

3.1 Experimental Overview

The main aim of this study was to gain insight into the mechanics of the FHL and FDL during locomotion. This was performed by utilizing the RDAS, a robotic apparatus that recreates the muscle activity and sagittal plane kinematics of lower extremity cadaver specimens during the stance phase of gait. Tendon forces, excursions, and forefoot plantar pressure distributions were measured under different toe flexor muscle control strategies in an attempt to investigate the likelihood that these muscles function isometrically in vivo.

3.2 Robotic Dynamic Activity Simulator

The RDAS is a second-generation device developed by Hoskins (2006) that is based on the preceding device known as the Dynamic Gait Simulator (Sharkey & Hamel 1998). The RDAS consists of a main chassis/frame and specimen carriage, which translates with respect to the main chassis (Figure 3-1). A force plate (Advanced Mechanical Technology Inc., Watertown, MA) is fixed to the frame, and serves as a surface for the simulations while providing measurements of ground reaction forces (GRFs) during the stance phase simulations. The RDAS can be broken down into three main subsystems: kinematics simulation, muscle simulation, and data acquisition. A brief description of these subsystems and overall RDAS functionality follows in the upcoming sections. In depth qualitative and quantitative analyses of the development of the RDAS as well as validation of its capabilities are described elsewhere (Sharkey & Hamel 1998, Hoskins 2006).
3.2.1 Kinematic Simulation

Three linear actuators (Industrial Devices Corporation, Salem, NH) rigidly fixed to the main frame, serve to translate the specimen carriage horizontally and vertically. A kinematic input profile database describing the sagittal plane positions of the lateral malleolus and fibular head for living subjects during normal locomotion is stored for access by the RDAS control software. This database provides time series positional input to a motor control board which controls the positions of the linear actuators. It includes positional data from a range of differently sized male and female subjects such that control parameters can be size-matched to individual cadaver extremities.
3.2.2 Muscle Simulation

A separate collection of six linear actuators (Industrial Devices Corporation, Salem, NH) serve to simulate the muscle activity of six lower extremity functional muscle units. Three muscles are controlled individually (flexor hallucis longus, flexor digitorum longus, and tibialis posterior) while the triceps surae (gastrocnemius & soleus), peronei (peroneus longus & peroneus brevis), and exensors (tibialis anterior, extensor digitorum longus, extensor hallucis longus & peroneus tertius) are each controlled as a group.

After stripping away the muscle tissue, clamps were fixed to the tendons of all six functional muscle units herein referred to as the clamp-tendon junctions (CTJs). The CTJ represents what would be the myotendinous junction in vivo. Small conduits located within the clamps allow for the routing of liquid nitrogen to the CJT to ensure a rigid fixation during simulations (Sharkey et al. 1995). Cables originating at each of the six linear actuators responsible for simulating muscle forces are coupled with a load cell, and fixed at the proximal end of each clamp. Therefore, each load cell measures the tensile force of the tendons of each functional muscle group. Also coupled to each cable is a displacement transducer, allowing for tendon excursion measurements during simulations.

The forces of each tendon are controlled by deriving target muscle force profiles. EMG profiles from all lower extremity muscles of interest were taken from Perry (1992), while PCSA data were taken from the work of Friederich & Brand (1990), which included a summary of Wickiewicz et al. (1983) and Weber (1846). Each of the six main functional muscle units were assumed to generate a force proportional to the ratio of its total PCSA to the PCSA of the Triceps Surae (TS). For the units comprised of more than one muscle tendon (i.e., Triceps Surae, Peronei, and Extensors), each muscle was assumed to generate a force proportional to its percentage of the unit’s total PCSA. Target force profiles are then derived under the assumption
that there is a simple linear relationship between EMG, PCSA, and muscle force. Each functional muscle unit is controlled by a microcontroller, which controls each tendon force based on feedback from the load cells (see Hoskins 2005 for more on the control system).

3.2.2 Data Acquisition

The RDAS and all its components are controlled by a personal computer with a LabWindows/CVI interface. The interface allows for the operator to select an appropriate kinematic input profile, adjust important parameters related to the simulation, display resultant GRFs, and calibrate and display muscle displacements and forces.

3.3 Specimen Preparation

Eight (see 5.3 Limitations of Study Design for comments on sample size) non-embalmed and non-paired fresh frozen lower extremity cadaver specimens were used in this study (Table 3-1). Specimens were thawed for at least twelve hours and inspected for any abnormalities and normal range of motion prior to preparation. All soft tissues proximal to the malleoli were removed, with care taken to preserve the full lengths of all tendons of the extrinsic foot muscles. The proximal end of the tibia was fixed to a coupling pot with the use of PMMA bone cement. The coupling pot was then secured to a metallic tibial fixator cup by a pin; the fixator cup serves as a rigid connection for the specimen to the specimen carriage of the RDAS.
Table 3-1. Information about the eight specimens used. Simulated body weight (BW) was selected based on the robustness of the specimen, and all GRF profiles and force measured to scaled by this value.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Side</th>
<th>Sex</th>
<th>Age</th>
<th>Simulated BW (kg, lbs)</th>
<th>Kinematic Profile</th>
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<td>M</td>
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<td>7_234_sub_2_trial_16</td>
</tr>
<tr>
<td>8</td>
<td>R</td>
<td>F</td>
<td>84</td>
<td>45 (100)</td>
<td>9_241_sub_5_trial_17</td>
</tr>
</tbody>
</table>

3.4 Experimental Protocol

3.4.1 Preliminary Validation Trials

Once secured into the RDAS, the specimen was subjected to a number of preliminary trials to ensure the simulation closely matched physiological loading during gait. To accomplish this, the RDAS allows for modulation of the simulation via two user-controlled inputs: initial height of the specimen carriage and TS force. After an appropriate kinematic input profile was selected, GRFs are recorded during a simulation and compared to target GRFs from the live subject database. Decreasing the specimen height allows for modulation of the first peak of the vertical GRF, while increasing the maximum TS force (and as a result, all functional muscle groups) allows for modulation of the second peak of the vertical GRF. These validation protocols ensured a close approximation of the stance phase of human walking.

3.4.1 Force-feedback Control Trials

Upon completion of the preliminary trials, three trials were performed at the ideal specimen carriage height and TS force. Tendon forces were visually inspected to ensure a close
match to target forces, ensuring fidelity of the force feedback circuitry. All tendon forces and
displacements were recorded, and these were referred to as force-feedback control (FC) trials.

3.4.2 Isometric Displacement Control Trials

Isometric displacement control (IDC) trials were designed to simulate isometric muscle function. Unlike FC trials designed to closely match target force profiles as derived from the EMG profiles of healthy subjects, PC trials are designed to maintain the CTJ in a stationary position and to record the tendon forces resulting thereof. The position at which to maintain the CTJ in IDC trials was determined through an iterative, trial and error process. To do this, an initial set-up trial was run with the CTJ positioned at the midpoint of the excursion range measured in the FC trials. The resultant force-time profile was then visually compared to the target profile. Based upon this comparison, the clamp location was repositioned proximally or distally by 1.0 mm to either increase or decrease the magnitude of the force carried in the tendon and a subsequent trial was executed. Set-up trials continued in this fashion until the best possible match was obtained between the measured and target tendon forces. Three trials were preformed for the FHL and FDL individually under IDC. Finally, three more trials were performed with both the FHL and FDL simultaneously under IDC (SIDC) at the same CTJ location used under individual IDC. All other muscle groups were maintained under FC for all experimental conditions.

3.4.3 Plantar Pressures

During all trials, the plantar pressure distribution of the specimen was recorded with a pedobarograph (EMed, Novel Scientific) fixed atop the force plate within the main frame of the RDAS. The plantar pressures were utilized to quantify forefoot loading. The variables of interest
obtained from the plantar pressure distributions were: contact area (cm²), maximum force (% of simulated BW), peak pressure (kPa), pressure-time integral (kPa*s), and force-time integral (% simulated BW*s) for each of six pre-determined regions of the forefoot. These pre-determined regions are computed based on the geometry of the foot in the default EMed software (Figure 3-2). This process is referred to as masking, and the Novel Standard Mask was applied to all simulated steps. The forefoot regions under investigation were the first metatarsal head (MTH1), second metatarsal head (MTH2), the combined third, fourth, and fifth metatarsal heads (MTH345), the first ray (Toe1), the second ray (Toe2), and the combined third, fourth, and fifth rays (Toe345). See Appendix A for EMed software manuals including masking definitions.
3.5 Data Processing

The RDAS provides displacement and force data during each simulation of stance phase. All of these data were filtered with a third order Butterworth filter and interpolated to 101 time points, each representative of 1% of stance phase. Forces were normalized to the body weight assumed in simulations, while all displacements were zeroed with respect to the initial positional
reading. Representative mean curves for each specimen and condition were calculated by averaging at each time point across the three trials.

The EMed software (see Appendix A), calculates the relevant plantar pressure variables throughout stance, separated by the desired masking regions of interest (i.e., the regions of the forefoot). The plantar pressure variables were averaged across the three trails for each specimen, plantar location and condition, and the grand means were calculated across specimens.

3.6 Statistical Analyses

Two tailed paired t-tests were performed to compare the mean tendon excursions of the IDC and SIDC trials to the FC condition. To determine differences between recorded and target tendon forces, 95% confidence intervals were constructed about the force-time curves. This process was repeated for displacement-time curves to compare excursions to neutral displacement, representing the myotendinous junction.

Target forces were regressed on actual tendon forces for each experimental condition as another method to compare the force-time curves. A perfect match of actual and target force curves would result in a regression y-intercept equal to 0, regression slope ($\beta_1$) equal to 1, standard error (SE) equal to 0, and a coefficient of determination ($R^2$) equal to 100%.

A general linear model ANOVA was performed with ‘specimen’ as the random factor and ‘experimental condition’ as the primary factor in order to compare plantar pressure variables. Each response variable for all six forefoot regions was analyzed to determine if experimental condition had a statistically significant effect ($\alpha = 0.05$). Power analyses were performed for each dependent variable to examine the statistical power of our sample.
Chapter 4

Results

4.1 Tendon Excursions

4.1.1 Force-feedback Control

FHL and FDL tendon excursions were recorded under FC and averaged across three trials for each specimen (Figure 4-1). The average FHL excursion range over all specimens was 7.18±1.75 mm, while the average FDL excursion range was 6.32±1.74 mm.

![Figure 4-1. FHL and FDL excursion range for each specimen under force feedback control, averaged over three trials.](image)

Most specimens exhibited a characteristic increase in displacement during the final 5-10% of stance, concurrent with toe-off which occurs at about 95% of stance in our simulations. High inter-specimen variability was evident (Figure 4-2, Figure 4-3); however, intra-specimen variability was generally low, indicating good reproducibility of our simulations. The high inter-
specimen variability is likely a function of anthropometry (e.g., toe length, foot length, tendon slack length), evidenced by a strong correlation ($r = 0.756, R^2 = 57.17\%$) between FHL excursion and FDL excursion (Figure 4-4).

Figure 4-2. FHL excursion range for each specimen under force feedback control, averaged over three trials. Note: positive excursion values are indicative of a distal displacement, representing simulated muscle lengthening.
Figure 4-3. FDL Tendon excursion for each specimen under force feedback control, averaged over three trials.

Figure 4-4. FHL excursion vs. FDL excursion, with line of best fit, for eight specimens under force feedback control ($r = 0.756, R^2 = 57.17\%$).
A representative mean displacement curve was created by averaging values at each instant across the eight specimens (Figure 4-5, Figure 4-6). The tendon excursion range for this cross-specimen average was 4.57±0.81 mm for the FHL and 5.13±1.21 mm for the FDL. Again, tendon shortening, indicated by an increase in tendon displacement, was most notable during the final 10% of stance. A large portion of the excursion range for the FHL did not deviate significantly from neutral position (i.e., 0 mm of displacement). For the FDL, approximately 50% of the tendon displacement curve was statistically different from neutral position.

Figure 4-5. Average FHL tendon excursion (±95% confidence interval) under force feedback control (n=8).
4.1.2 Isometric Displacement Control

Individual trials were performed with just the FHL under position feedback control (FHL IDC) and with just the FDL under position feedback control (FDL IDC; Figure 4-7). The average range of FHL and FDL tendon excursions under individual IDC were 2.33±0.58 mm and 1.45±0.45 mm, respectively. Under individual IDC, the excursion ranges of both the FHL and FDL decreased significantly with respect to FC trials (p < 0.001). The excursion ranges under simultaneous isometric displacement control (SIDC) were also significantly lower (p < 0.001) for both the FHL (2.75±0.73 mm) and FDL (1.79±0.49 mm) as compared to FC (Figure 4-8). The excursion ranges under IDC were statistically different than natural position. However, the
ranges were relatively small, and all ranges differed from neutral in the same systematic way due to limitations of the RDAS hardware.

Figure 4-7. Comparison of mean tendon excursion (± 1 SD) for FHL and FDL across experimental conditions (n=8). Star denotes a significant (p<0.001) difference of IDC (isometric displacement control) or SIDC (simultaneous IDC) with respect to FC (force feedback control).
Figure 4-8. Tendon excursions (±95% confidence interval) under isometric displacement control (individual and simultaneous) tendon excursions.

4.2 Tendon Forces

4.2.1 Force-feedback Control

Average force profiles across all three FC trials (actual force) were constructed and compared to desired force profiles derived from EMG patterns (target force). FHL tendon forces under FC peaked at approximately 22% of BW, by design. Force profiles under FC closely matched target forces (Figure 4-9) illustrating high fidelity of the force feedback control system of the simulator. As a result of this, the target force profiles can serve as a control for tendon forces, and plantar pressure distributions under FC can serve as a control for comparisons of forefoot mechanics.
FDL tendon forces under FC peaked at approximately 6.5% of BW, by design. Although the FDL force profile closely matched the pattern of the target profile, the actual force was statistically different compared to the target force profile (Figure 4-10). This indicates that the force feedback control was not as precise for the FDL as it was for the FHL. Although statistically different, it is assumed that the force profile under FC is close enough to regard it as a control, in a similar fashion to the FHL.
4.2.2 Isometric Displacement Control

The individual IDC tendon forces followed a similar pattern to that of the target force profile (Figure 4-11). The FHL tendon force under IDC was not statistically different for a majority of the simulation, excluding a brief period at 50% stance. The FDL force did exhibit a statistically significant difference for almost the entirety of stance. However, the general patterns of IDC tendon force were similar to target forces. In addition, the peak force values, and time to peak force values were also similar. The SIDC trials exhibited a similar pattern to that of individual IDC (Figure 4-12).
Figure 4-11. FHL and FDL tendon forces under individual isometric displacement control compared to target forces.
Regression analyses were performed in an attempt to quantify the similarities between target forces and actual forces for each experimental condition (Table 4-1, Table 4-2). Comparisons were made for each specimen individually, and for the representative mean (ALL). Under FC, FHL forces were very close to target forces, evidenced by a high coefficient of determination (99.9%) and slope close to 1 (1.024). Under IDC and SIDC, the correlations remained very strong ($R^2 = 95.8\%$ and 93.2%; $\beta_1 = 0.919$ and 0.844, respectively). FDL tendon forces also exhibited a very strong correlation under FC ($R^2 = 98.8\%$, $\beta_1 = 0.956$). The correlations were still relatively high under IDC and SIDC ($R^2 = 75.4\%$ and 68.6%; $\beta_1 = 0.904$ and 0.838). A representative specimen (Specimen #2) is presented in Figure 4-13 and Figure 4-14, with the corresponding regression analyses.
Figure 4-13. Regression analyses of actual force versus target force for the FHL under FC, IDC, and SIDC for Specimen #2.

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<tr>
<th>Spec</th>
<th>β₀</th>
<th>β₁</th>
<th>SE</th>
<th>R²</th>
<th>β₀</th>
<th>β₁</th>
<th>SE</th>
<th>R²</th>
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<td>2.051</td>
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Table 4-1. Results of regressing FHL target force on actual force for each experimental condition. β₀ is the y-intercept of the regression line, β₁ is the slope, SE is the standard error of the residuals, and R² is the coefficient of determination. ‘ALL’ represents the representative mean force-tendon curve.
Figure 4-14. Regression analyses of actual force versus target force for the FDL under FC, IDC, and SIDC for Specimen #2.

Table 4-2. Results of regressing FDL target force on actual force for each experimental condition. $\beta_0$ is the y-intercept of the regression line, $\beta_1$ is the slope, SE is the standard error of the residuals, and $R^2$ is the coefficient of determination. ‘ALL’ represents the representative mean force-tendon curve.
4.3 Plantar Pressures

Plantar pressure distributions were recorded across the forefoot for all specimens for each of the four experimental conditions. Under FC, contact areas for the MTH345 region were highest (23.48±4.98 cm²), while they were lowest for the Toe2 region (4.81±1.44 cm²), reflective of the relative size of each mask region of interest (Figure 4-15). Contact area means were similar across experimental conditions.

Under force control, maximum forces under the first metatarsal head were about 20% of BW, while forces under the first ray were about 25% of BW (Figure 4-16). The second metatarsal head exhibited maximum forces of around 30% of BW, while the second ray showed much smaller maximum forces compared to the first ray (approximately 5% of BW). The values of maximum force were similar for each region, regardless of experimental condition.
Figure 4-16. Maximum force (±1 SE) for each region of the foot across experimental conditions.

Peak pressures ranged from approximately 500 kPa (MTH345, MTH2, Toe1) to 100 kPa (Toe345). It is interesting to note that the peak pressures were much higher in Toe1 compared to MTH1, but the peak pressures in Toe2 were much lower compared to MTH2 (Figure 4-17). For all regions, the peak pressures did not appear to change significantly as a function of experimental condition.

Figure 4-17. Peak pressures (±1 SE) for each region of the foot across experimental conditions.
In addition to peak pressures and forces, pressure-time and force-time integrals were calculated to examine if there were any differences over time between experimental conditions. Under FC, pressure-time integrals ranged from approximately 2250 kPa*s (MTH345) to 500 kPa*s (Toe345; Figure 4-18). Force time intervals were markedly higher for MTH345 when compared to the other regions (Figure 4-19). This can be attributed to the masking definition protocols (MTH345 represents a much higher contact area with respect to the other regions, see Figure 4-15).

![Figure 4-18. Pressure time integrals (±1 SE) for all mask regions across experimental conditions.](image)
4.3.1 Statistical Analyses

Analysis of variance tests confirmed that experimental condition did not have any significant effects on any variables of interest (Table 4-3).

<table>
<thead>
<tr>
<th></th>
<th>Contact Area</th>
<th>Maximum Force</th>
<th>Peak Pressure</th>
<th>Pressure Time Integral</th>
<th>Force Time Integral</th>
</tr>
</thead>
<tbody>
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<td>0.151</td>
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<tr>
<td>MTH 345</td>
<td>0.359</td>
<td>0.89</td>
<td>0.077</td>
<td>0.641</td>
<td>0.506</td>
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</tbody>
</table>

Table 4-3. General linear model ANOVA p-values for each response variable and each region of the foot.

Power analyses performed on each variable ranged from 0.0753 to 0.5373, all well below the commonly accepted value of 0.8 (Table 4-4). This indicates that our sample is underpowered, preventing us from stating conclusively that there are no significant differences between experimental conditions.
<table>
<thead>
<tr>
<th></th>
<th>Contact Area</th>
<th>Maximum Force</th>
<th>Peak Pressure</th>
<th>Pressure Time Integral</th>
<th>Force Time Integral</th>
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<td>MTH2</td>
<td>0.078</td>
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<td>0.135</td>
<td>0.276</td>
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<td>Toe 2</td>
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<td>Toe 345</td>
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<td>0.083</td>
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Table 4-4. Statistical powers for each response variable and each region of the foot.
Chapter 5

Discussion

5.1 Hypotheses

Hypothesis 1: The FHL functions isometrically during the stance phase of gait, with the geometry and kinematics of the foot dictating functionally appropriate tendon force.

Average FHL tendon excursion under FC was relatively small (7.18±1.75 mm), or approximately 13.6% of the optimal muscle fiber length (Ward et al. 2009). This excursion is representative of the change in position of the myotendinous junction in vivo, which is a function of the change in length of the distal tendon, muscle fibers, and proximal tendon. Although inter-specimen variability of excursion over stance was fairly high, most specimens followed the same characteristic pattern. There was a distal displacement, indicative of muscle lengthening, briefly after heel strike (at approximately 20% of stance), and again around toe-off (at approximately 90% of stance). For most specimens, there were no displacements in either direction of more than approximately ±2 mm from neutral position throughout the majority of stance. Average excursions for all specimens were not significantly different than zero during more than one-half of stance phase (Figure 4-5).

There were no significant differences in FHL tendon forces between FC trials and target force profiles derived from EMG. Under IDC, the average force-time curve was similar in shape to target forces over a majority of stance. This indicates that functionally relevant, dynamic tendon loads can occur during isometric contractions of FHL. Since isometric control is
independent of force \((i.e., \text{the controller only seeks to maintain constant length})\), tendon force dynamics are solely a function of the kinematics of the foot and shank during stance.

Regressions of target forces on actual forces for the experimental conditions were used as an additional method to quantify the similarities between the control strategies. The linear regression of target FHL force on FC FHL force resulted in a very strong correlation \((R^2 = 99.9\%, \beta_1 = 1.024)\). Under IDC, the correlations remained very strong \((R^2 = 95.8\%, \beta_1 = 0.919)\). This serves as additional evidence that the force profiles were not drastically different between conditions, an indication that physiologically realistic FHL tendon loading is possible during isometric functioning. Taken as a whole, evidence gathered here suggests that this muscle is functioning nearly isometrically \textit{in vivo}.

ANOVA output confirms that no plantar pressure variables exhibited a statistically significant change during IDC compared to FC for all regions of the forefoot. Especially noteworthy is the lack of change in the MTH1 and Toe1, regions that would presumably be most affected by aberrant FHL function. We could not detect any change in forefoot mechanics under an isometrically functioning FHL.

FHL tendon forces and forefoot loading mechanics are not remarkably different regardless of the FHL muscle control strategy employed. Therefore, it is concluded that the FHL may be functioning isometrically \textit{in vivo}, while the force in the tendon is a function of the kinematics of the lower shank during stance and the anthropometry of the foot. This conclusion is consistent with that of previous work (Kirane \textit{et al.} 2008).

\textit{Hypothesis 2: The FDL does not function isometrically during the stance phase of gait; using this control strategy will produce functionally inappropriate tendon forces.}
Mean FDL tendon excursion was limited to a narrow range during FC (6.32±1.74 mm), or approximately 14.2% of optimal muscle fiber length (Ward et al. 2009). As noted above, this excursion is representative of the change in location of the myotendinous junction in vivo. Under FC, the FDL tendon excursion followed a slightly different trend compared to the FHL. Significant excursion did not occur until approximately 25% of stance, where there was a small proximal displacement (suggestive of concentric muscle contraction) up to terminal stance. At this point, excursion reversed (i.e., the simulated muscle lengthened) until toe-off. Despite this dissimilar timing, overall FDL excursion under FC was similar to the FHL in that, when averaged across specimens, it appeared to deviate little from neutral position. We measured proximal displacement (i.e., shortening) early in our simulations during FC, possibly indicative of concentric contraction. However, the overall excursion was in a very narrow range, suggesting that near isometric contraction may occur during stance in vivo, contrary to our hypothesis.

FDL tendon force under FC followed a pattern very similar to that of target forces, calculated from EMG, although forces were consistently and significantly lower throughout the simulation. This is a potential limitation of the RDAS hardware, where the force-feedback control loses fidelity at lower loads. Despite this limitation, there was still less than 1% BW difference between FC and target force profiles. Thus, it is concluded that FC trials are an adequate representation of target force profiles.

FDL tendon forces under IDC followed a pattern very similar to that of the target forces, although actual forces were statistically greater for most of stance, in partial support of our hypothesis. Despite these differences, peak tendon force and time to peak tendon force were similar. Regressing target force on actual forces under FC resulted in a very strong correlation ($R^2 = 98.8\%$, $\beta_1 = 0.956$). Additionally, regressing target force on the IDC force profile resulted in a good correlation ($R^2 = 75.4\%$, $\beta_1 = 0.904$), although not as strong as the FHL under IDC. Based on the evidence from linear regression, the similar characteristic patterns, peak values, and
time to peak values, it seems reasonable to conclude that FDL forces under IDC are not remarkably different than target forces derived from EMG data. It is concluded based on this evidence that, contrary to our hypothesis, physiologic and functionally appropriate loading in the FDL tendon is possible under isometric function of the FDL.

The FDL tendon inserts on the distal phalanges of the 2-5th rays of the foot. Therefore, if an isometrically functioning FDL muscle significantly altered forefoot mechanics, one would expect changes in plantar pressure variables of the MTH2, MTH345, Toe2, or Toe345 regions. However, there were no statistically significant differences between IDC and FC conditions for any plantar pressure variable. This indicates that forefoot mechanics are not significantly altered under simulated isometric contraction of the FDL.

In summary, it appears that functionally appropriate loading in the FDL tendon is possible under simulated isometric function of the FDL. Additionally, there was no change in forefoot loading under an isometrically functioning FDL. Therefore, it seems plausible to conclude that, contrary to our hypothesis, that the FDL may indeed be functioning isometrically in vivo.

Hypothesis 3: Simultaneous isometric control of the FDL and FHL during the stance phase of gait will result in a significant increase in tendon forces and forefoot loading.

There is evidence from the individual IDC trials that the FDL and FHL may be functioning isometrically during the stance phase of gait. Trials were also performed in which both FHL and FDL were controlled isometrically at the same time (SIDC). FHL and FDL tendon forces under SIDC were similar to those measured during individual IDC. Under both of these conditions, the FHL tendon force was closer to its target force profile than the FDL was to its
respective target force profile. Peak force and time to peak force were similar for the both the FHL and FDL when comparing target force to SIDC force.

ANOVA output again confirms that there are no significant differences for any plantar pressure variable across any region of the forefoot during SIDC compared to FC. This provides evidence that, contrary to our hypothesis, forefoot mechanics are not significantly altered during simultaneous isometric function of both toe flexor muscles. Although other plantarflexor muscles have been shown to have a significant effect on balancing forefoot loading (Ferris et al. 1995), the effects of their simulated isometric functioning on forefoot mechanics is unknown.

This study provides evidence that isometric functioning of the toe flexor muscles during the stance phase of gait produces physiologically realistic tendon forces and forefoot loading. From this evidence, it is a reasonable to suggest that these muscles may be functioning isometrically in vivo during stance phase in human locomotion.

5.2 Implications

The FHL and FDL have been shown to have a significant effect on loading of the forefoot (Ferris et al. 1995), and altered function of these muscles can lead to gait abnormalities (e.g., Feeney et al. 2000). Therefore, a comprehensive knowledge of the function of these muscles during normal habitual locomotion can be important in the detection and prevention of foot pathologies. For example, if these muscles indeed function isometrically during stance, then any injury or pathology that alters the position of the muscle or tendon length might have critical effects on normal ambulation.

Analysis of the function of multi-articular muscles has been performed in the past using inverse dynamics (e.g., Prilutsky and Zatsiorsky 1994), EMG readings (e.g., van Ingen Schenau 1987), and forward dynamic simulations (e.g., Jonkers et al. 2003, Neptune et al. 2004). To date,
there is little direct experimental evidence examining the mechanics of multi-articular muscles in vivo or in situ. More specifically, there have been few studies examining the function of the multi-articular FHL and FDL during tasks such as locomotion. Our high fidelity in situ model loads cadaver lower extremities under conditions very close to those occurring in life during the stance phase of gait, thereby providing an ideal method with which to analyze the function of these extrinsic foot muscles.

Multi-articular muscles are hypothesized to function isometrically in order to redistribute segmental energy in ways that might not be possible with just mono-articular muscles (Zajac et al. 2002). The results of the present study are consistent with this hypothesis. These muscles, and all multi-articular muscles by definition, simultaneously contribute to the motion of numerous lower extremity joints critical to normal locomotion. Zajac et al. (2002) posit that the challenge with respect to the analysis of multi-articular muscles is to determine what mechanical, metabolic, or neural control benefits they provide. Therefore, these topics will be explored in further detail with respect to the FHL and FDL.

The magnitude and speed of a muscle’s force production is dependent on whether the muscle fibers are shortening (concentric), lengthening (eccentric), or not changing (isometric) during contraction. All of these forms of muscle contraction have an associated metabolic cost, i.e., energy expenditure. Studies that have examined the metabolic cost of muscle function suggest that generating force requires more energy than maintaining force (Russ et al. 2002, Newham et al. 1995, Beltman et al. 2004). Thus, isometric contraction likely has an associated decrease in metabolic cost when compared to lengthening or shortening contractions. Fukunaga et al. (2001) demonstrated that the human gastrocnemius functions isometrically during stance, possibly to ensure the muscle is operating in the optimal regions of its force-velocity and force-length curves. The authors suggest this function also allows for elastic strain energy return from the Achilles tendon during toe-off. The FHL and FDL supplement the plantarflexion actions of
the gastrocnemius and, as such, may operate under the same energy conserving principles during normal locomotion. It could in fact be that all distal multi-articular muscles of the lower extremity function isometrically during locomotion in vivo. Certainly these muscles function under both concentric and eccentric conditions in other tasks that require forceful independent flexion or extension (Reeser et al. 1983). For habitual tasks like locomotion, storage of strain energy in the tendons may be the most energetically efficient way to produce the tendon forces necessary to maintain normal locomotion.

The role of the CNS in controlling locomotion is a widely researched topic; the results from this study suggest that the CNS is likely controlling the FHL and FDL as a spring, in an attempt to maintain a constant sarcomere length. A length-servo mechanism such as the stretch reflex is a likely control strategy for these two muscles during ambulation. This reflex may be triggered when muscle spindles are stretched during dorsiflexion of the talocrural joint at midstance, resulting in an increase in muscle excitation evident from EMG patterns (Perry 1992). If these muscles are indeed controlled via a stretch reflex mechanism, there might be implications for the study of peripheral neuropathies (e.g., spastic paresis), or diseases in which the stretch reflex is impaired (e.g., diabetes).

There is evidence that collections of neurons within the mammalian spinal cord (the central pattern generator, CPG) contribute to the rhythmic neural excitation pattern necessary to maintain normal locomotion (Goulding 2009; Guertin 2009; Grillner 1985; Minassian et al. 2007). These results may provide another basis with which to describe the role of the CPG in controlling locomotion; perhaps the CPG controls proximal muscles, while something such as the stretch reflex provides local length control of distal muscles. If the FHL and FDL are controlled by a stretch reflex mechanism, then the CNS would only have to expend processing power on tasks that require more forceful flexion of the toes (e.g., running, jumping). The intrinsic musculature of the foot might serve as a means to modulate the function of the isometrically
controlled long toe flexors. Previous studies examining locomotion in decerebrate cats suggest that there is differentiated neural drive to the FHL and FDL (Burke et al. 1985, O'Donnovan et al. 1982). However, our results suggest that, in humans, these muscles are likely functioning in the same fashion, indicating that they may be controlled in the same fashion. The contrast in these findings is likely due to the drastic differences in anatomy and style of locomotion between man and other animals; hence, the differences in control strategies may be a result of the evolution of bipedalism in humans.

Muscles are thought to be uniquely suited to function through their architecture including, but not limited to: pennation angle, fiber length, and fiber orientation. The FHL and FDL have rather different muscle masses of 38.9 kg and 20.3 kg, respectively. In addition, the FHL has a higher pennation angle (16.9º vs. 13.6º) and greater PCSA (7.2 cm² vs. 4.5 cm²; Ward et al. 2009). Muscle architecture is a better predictor of muscle force production than muscle mass or volume (Lieber et al. 2011). Although muscle fiber lengths are relatively different in these two muscles (FHL = 52.7±12.9 mm, FDL = 44.6±10.6 mm), the ratio of fiber length to muscle length is similar (FHL = 0.16, FDL = 0.20; Ward et al. 2009). A ratio close to one would indicate a higher tendency toward shortening during contraction (e.g., the sartorius with a ratio of 0.90). Relatively low values of the fiber length to muscle length ratio are indicative of muscles that may potentially function isometrically during stance. It is interesting to note that the peroneus longus, peroneus brevis and tibialis posterior also exhibit similar low ratios, as does the gastrocnemius, which has been demonstrated to function isometrically during walking and jumping (van Ingen Schenau et al. 1989, Fukunaga et al. 2001, Prilutsky & Zatsiorsky 1994).

A more comprehensive understanding of how the FHL and FDL meet the high functional demands of human gait can contribute to a better understanding of consequences that would arise as a result of their compromised function. For example, a recent study in our laboratory has shown that tenosynovitis at the myotendinous junction of the FHL increases loading of the first
MTP joint and may play a role in the generation of hallux rigidus (Kirane et al. 2008b). Thus, the position at which the myotendinous junction is held stationary is critical to producing proper loading in the FHL tendon. Extending this finding, it becomes clear that any surgical intervention affecting the length, displacement, or altering the anatomy/geometry of these tendons may disrupt normal function and significantly compromise efficient locomotion.

5.3 Limitations of Study Design

The primary limitations from the present study are related to the current capabilities of the RDAS. A simple linear relationship between muscle force and EMG was assumed in order to construct target force profiles with which the RDAS controlled the force in all muscle groups during the force-feedback control trials. The RDAS cannot currently simulate intrinsic muscle function, which has been shown to be active during gait and may have a significant effect on the force output of these tendons (Reeser et al. 1983). The RDAS currently simulates the time period of stance at a 20:1 ratio (i.e., typically stance occurs for 0.6 seconds, and it lasts for approximately 12 seconds during our simulations). It was assumed that viscoelastic effects on tendons and other soft tissues were minimal.

Under isometric-displacement control, tendon excursions were significantly smaller compared to FC; however, these excursions did deviate somewhat from neutral position. This indicates that there are some limitations in the displacement control system of the RDAS. The excursion range under isometric-displacement control did vary in the same systematic way across trials (Figure 4-8).

The primary limitation of our experimental protocol is low sample size, preventing us from conclusively stating that there are no significant differences between plantar pressure variables of interest across conditions. The means were very similar however, thus making it
difficult to have a power of 0.8 without an extremely high number of samples. There may be unknown order effects on our results since our protocol was always performed in the same sequence (FC, FHL IDC, FDL IDC, and S IDC). This was unavoidable due to the need to perform the FC trails first, in order to determine the proper isometric position at which to hold the tendon during IDC. It was also necessary to perform the S IDC trials last, after the appropriate isometric positions were determined for the FHL and FDL.

5.4 Future Work

Future work could examine the functioning of other extrinsic muscles of the foot (e.g., peroneus longus and brevis). It may turn out that many of these muscles function isometrically, and the force in the tendon is modulated by the kinematics of the foot and ankle during a task, and by intrinsic foot muscles. Recording bone strains in the metatarsals would be a more direct way to quantify forefoot loading, specifically metatarsophalangeal JRFs during simulations.

It would be interesting to examine the potential isometric function of the toe flexors during other activities, such as running. Perhaps these and all extrinsic muscles function very differently depending on strike pattern mechanics during running (i.e., forefoot or heel striking). To drive the RDAS, kinematic profiles would need to be collected during running under different strike patterns. In addition, EMG recordings from indwelling electrodes would be necessary in order to re-create target force profiles appropriate for running. This study would provide an opportunity to record muscle forces and bone strains in the foot, something nearly impossible to do in vivo. These results could provide interesting input to the current barefoot running debate (e.g., Lieberman et al. 2010).
5.5 Concluding Remarks

This study provides experimental evidence from actual anatomical samples that the behavior of the forefoot is not mechanically altered under isometric control of the extrinsic toe flexor muscles during stance. It seems likely that an isometric control strategy is used in vivo to minimize the computational and energetic cost of locomotion. To date, few studies have been performed to examine the mechanics of the extrinsic toe flexor muscles during ambulation. These muscles, the FHL in particular, experience very high loads during stance, and play an important role in regulating forefoot mechanics. Our data suggest that these muscles are functioning isometrically and likely under the control of a length-servo mechanism such as the stretch reflex.
References


flexors on forefoot loading during heel rise. *Foot Ankle Int* **16**: 464-73.


Appendix A

EMed Software Manuals
3. THE PROGRAM automask

This program is used to automatically create standard masks for a footprint. A mask is a region of interest overlaid onto a pressure picture. Several masks are available for quick and easy masking of a variety of foot regions. The masks are created for data analysis in the programs multimask evaluation and groupmask evaluation from the novel scientific software. More than one data and mask file can be opened with this program.

3.1 Working with automask

There are three options for creating masks in this program.
1. Select a predefined mask for the entire foot.
2. Select a predefined mask for one region.
3. Create a new mask.

3.1.1 Select a Predefined Mask for the Entire Foot

Select 'mask' from the menu and a list of the available automatic masks will be shown. Click on the desired mask. The masks are as follows:

- novel mask

  This mask separates the foot into heel, midfoot, forefoot (consisting of 1\textsuperscript{st}, 2\textsuperscript{nd}, 3\textsuperscript{rd}, 4\textsuperscript{th} and 5\textsuperscript{th} metatarsal heads), hallux, second toe and the lateral toes.

  The boundary between the heel and the midfoot is defined by default as 73\% of the foot length from the toes to the heel. The boundary between the midfoot and the forefoot is defined by default as 45\% of the foot length from the toes to the heel. The boundary between the forefoot and toes and also between toes is defined by taking into consideration the values of peak pressure under the toes and the gradients of pressure around these maximum values.

  The angles that define the metatarsal heads are given in
percent of the long plantar angle. Drawing tangents for the medial and lateral sides of the maximum pressure picture produce this angle. The tangents meet to form the long plantar angle. From the medial to the lateral side of this angle the metatarsal heads are defined as being 30%, 17%, 17%, 17% and 19% of the long plantar angle.

Taken from the Cavanagh and Ulbrecht (1994) mask for the diabetic foot, this mask divides the foot into lateral and medial heel, lateral and medial midfoot, forefoot (consisting of 1\textsuperscript{st}, 2\textsuperscript{nd} and lateral metatarsals heads), hallux, 2\textsuperscript{nd} toe and lateral toes.

The boundaries between the heel and midfoot and the midfoot and forefoot are defined by default as 73% and 45% of the foot length from the toes to the heel respectively.

The lateral and medial sections of the heel and the midfoot are defined by a foot axis that is drawn from the centre of the heel to the centre of the second toe.

The 1\textsuperscript{st}, 2\textsuperscript{nd} and lateral metatarsals heads are separated by straight lines which are drawn parallel to the foot axis and which divide the forefoot region vertically from the medial to the lateral side into sections of 30%, 25% and 45% respectively. (Hernig, 1984)

This mask divides the maximum pressure picture into heel, midfoot, forefoot and toes.

The heel to midfoot and midfoot to forefoot divisions are defined (by default) using 73% and 45% of the foot length from toes to heel respectively.

The forefoot to toes division is defined using pressure gradients around the peak pressures of the toes.

This mask divides the maximum pressure picture into heel, midfoot, forefoot, hallux and lateral toes. The heel to midfoot and midfoot to forefoot divisions are defined (by default) using 73% and 45% of the foot length from toes to heel respectively.

The forefoot to toes division and the between toes division are defined using pressure gradients around the peak pressures of the toes.
- **6 mask standard division**

  This mask divides the maximum pressure picture into heel, midfoot, forefoot, hallux, 2nd toe and lateral toes.

  The heel to midfoot and midfoot to forefoot divisions are defined (by default) using 73% and 45% of the foot length from toes to heel respectively.

  The forefoot to toes division and the between toes division are defined using pressure gradients around the peak pressures of the toes.

- **4 mask rectangular division**

  This mask again divides the maximum pressure picture into heel, midfoot, forefoot and toes.

  The heel to midfoot and midfoot to forefoot divisions are defined using 73% and 45% of the foot length from toes to heel respectively.

  Different algorithms than those used for the standard masks are used to distinguish the toes from the forefoot. These algorithms use fewer points than those for the other masks. A rectangle is defined parallel to the foot axis. This axis is drawn using a least squares method. The boundaries between the heel, midfoot and forefoot regions are perpendicular to this axis.

  The main advantage of using this mask is that the gait line or COP is still recorded even if it travels outside the area of the maximum pressure picture. This is also the mask used in the program regional velocity of the COP.

- **11 mask standard division**

  This mask is exactly the same as novel mask except that the heel is divided into medial and lateral sections by a line joining the center of the heel to the center of the second toe.

These masks can be edited as desired.

1. Open the data file.
2. Select a predefined mask file.
3. Change mask by using percentages.
   a. Click the options icon. Several divisions of each automatic mask can be edited using the 'options' menu. The following screen will appear:
b. Change the percentage of the section by typing in the new value.

NOTE: The divisions above are used in several of the automatic masks and may be altered by simply typing in new values. If changing the percentage division of the metatarsal heads in a mask that uses this method, then the section of the fifth metatarsal head is always the remaining % from a total of 100.

c. Select OK and the masks will be created using the new divisions.

d. Select Default to return to the default settings shown above.

   a. See section 3.1.3

3.1.2 Select a Predefined Mask for One Region.
   1. Select the 'Foot regions' menu.
   2. Select any of the regions from the list.

<table>
<thead>
<tr>
<th>Pixels</th>
<th>Predefined Mask</th>
</tr>
</thead>
<tbody>
<tr>
<td>Big Toe</td>
<td>pic Big Toe</td>
</tr>
<tr>
<td>Toe 1</td>
<td>pic Toe 1</td>
</tr>
<tr>
<td>Toe 2</td>
<td>pic Toe 2</td>
</tr>
<tr>
<td>Toe 3</td>
<td>pic Toe 3</td>
</tr>
<tr>
<td>Toe 4</td>
<td>pic Toe 4</td>
</tr>
<tr>
<td>Toe 5</td>
<td>pic Toe 5</td>
</tr>
<tr>
<td>Toes 2345</td>
<td>pic Toes 2345</td>
</tr>
<tr>
<td>Toes 345</td>
<td>pic Toes 345</td>
</tr>
<tr>
<td>Toes 45</td>
<td>pic Toes 45</td>
</tr>
<tr>
<td>Novel Metatarsal 1</td>
<td>pic Novel Metatarsal 1</td>
</tr>
<tr>
<td>Novel Metatarsal 2</td>
<td>pic Novel Metatarsal 2</td>
</tr>
<tr>
<td>Novel Metatarsal 3</td>
<td>pic Novel Metatarsal 3</td>
</tr>
<tr>
<td>Novel Metatarsal 4</td>
<td>pic Novel Metatarsal 4</td>
</tr>
<tr>
<td>Novel Metatarsal 5</td>
<td>pic Novel Metatarsal 5</td>
</tr>
</tbody>
</table>

The appropriate regions will be masked. These regions are part of the full foot masks listed in the previous section.
NOTE: This option can be helpful for those less than 'normal' feet. The computer may easily be able to identify part of the foot, say the heel, but have difficulties with the toes. With this method, the heel can quickly be identified and then a manual mask can be drawn for the toes.

3. Change mask manually using toolbar.
   See section 3.1.3 for editing and creating masks.

3.1.3 Create and Edit Masks

New masks can be created manually over an empty footprint or one that already contains automatically generated masks. When these masks are saved and reopened for another data file they are oriented so that they fit the new footprint whether it was left or right.

1. Open one or more data files
2. Draw up to 32 masks in the active window (blue title line).
   a. To draw the masks, use the buttons in the tool bar at the left side of the program window or alternatively use the commands in the Mask menu.

New
Click on this button and the mouse tip changes to a circle in the active window. Close to the small circle there is a square, indicating that a new mask can be drawn. Place the mouse over the intended starting point for the mask and click the left mouse button. Drag and click, to set further points, creating the mask. To close the mask just after setting the last point, click on the right mouse button.

Delete
This button is used to delete a mask. Click on the button and the mouse tip becomes a cross. Place the cross on the mask to be deleted and click the mouse button.

Change
To modify the size and shape of a mask, use this button. The mouse tip looks like a pair of pincers and the corner points of the mask can be moved with it. When creating the mask, the points at which the mouse was clicked, a node was created. Click on the nodes to move the lines, or click in the middle of a line to create a new node.

Mirror
With this command all masks in the active window will be mirrored by a horizontal or vertical line.

Copy
Masks can be copied from one data file to another. Open both data files and arrange them next to each other (use the function Tile in the Windows menu). Then activate the window containing the masks, click on Copy button and click on the masks and drag them to the other window. Release the mouse button and the masks are copied.
Move single mask
Click on this button to shift one mask on the pressure picture. Click on and drag the mask to the new location.

Move all masks
This function allows the entire mask arrangement to be moved within the window. Click anywhere in the window and drag masks to the new location.

Mask Number
The number and the name of masks can be changed using this button. Activate the function and click on the mask line to be edited: The following dialog box will appear to allow the changes.

NOTES:
- The name and the number of the mask are displayed in the status line at the bottom of the window as soon as the mouse tip is close to the edge of the mask.
- A sensor belongs to a mask if its center lies within the mask. Masks can overlap or lie within another mask.
- **Copy masks to clipboard** can be used in the same way as the **Copy** button of the toolbar to copy a mask to the clipboard. **Paste masks from clipboard** inserts a mask arrangement from the clipboard into the pressure picture. Note however that contrary to the programs *automask* (in *novel-ortho*) and *creation of percent masks* here the masks are not adjusted to the size of the pressure picture.
- To undo the previous action, click on **Undo** in the **Edit** menu.

3. Save the created masks
   a. Click **File**, **Mask save** or **Save all masks**.
      **NOTE:** Mask save will only save one mask file at a time. Save all masks will automatically save all masks opened over a data file using the same name as the data file. If a different name is desired, the masks must be saved individually.
   b. There are three file types for which mask files can be saved as.
      i. *.*msa – absolute mask: Mask is saved relative to the platform. Mask size is not adjusted to fit different sized pressure pictures and is not shifted to fit over the pressure picture.
      ii. *.*msr – relative mask: Mask is saved relative to the loaded rectangle. Mask does not change size, but shifts to fit over the pressure picture.
      iii. *.*std – automask: The mask is adjusted to fit over the pressure picture.

4. Close the program.
21. **THE PROGRAM groupmask evaluation**

This program is used to evaluate and statistically analyse dynamic pressure related variables, in different regions of interest (masks), for a group of files. The group of files must have been previously created in **group editor** as a *.par file.

### 21.1 Working with **groupmask evaluation**

1. Start the program
2. Open the group file (*.par)
   a. Click File/Group open. An **Open data** window appears.
   b. Select a group file to open. The following screen will appear.

![Open Data Window](image)

   c. The files can be normalized to body weight if the option is selected. Body weight must have been entered in the group file. A risk threshold can be set here.

**NOTE:** For pedar files only, select rather to analyse data over each step in the file (**Statistics over steps in trials**) or just over the entire file (**Statistics over trials**).

If the file cannot be opened, the following warning will appear. Most likely the file path is no longer correct.

![Warning Window](image)

**NOTE:** A warning message will not appear if the matching mask file cannot be found! Make sure the data files and mask files all fit together as desired.

3. After loading the data files the following window will appear:
4. Select the parameters to be evaluated by clicking on the checkboxes in the list. Confirm the selection with OK. The selected parameters can be changed later by opening this window with Select Pictures in the Pictures menu.

21.2 Analysed Parameters

After choosing the parameters to evaluate, the Pressure picture with masks will appear. Verify the masks and file fit as desired.

The other parameters are displayed using bar and line graphs (showing time dependencies). In general, pressure data are shown in blue, force data in green, area data in red, and time data in brown. The following parameters can be analysed:

21.2.1 Bar graphs

General Information:
- For each mask, the bar shows the mean value of the parameter with the standard deviation (or 95% confidence interval) in yellow.
- The numbers above each bar represent the value of the parameter and the standard deviation.
- The results for the whole object are listed at the top of the graph.

- Maximum force
  The mean value of the maximum force that occurred in each measurement for each mask.
- Instant of maximum force
  The average time the maximum force occurred in each measurement (measured in ms).
- Instant of maximum force [%ROP]
  The average time the maximum force occurred in each measurement (measured in %ROP).
- Peak pressure
  The mean value of the peak pressure that occurred in each measurement for each mask.
• **Instant of peak pressure**
The average time the peak pressure occurred in each measurement (measured in ms).

• **Instant of peak pressure [\%ROP]**
The average time the peak pressure occurred in each measurement (measured in \%ROP).

• **Mean Force**
The average of the total mean force in each measurement for each mask.

• **Mean Area**
The average loaded area in each measurement for each mask.

• **Maximum Mean Pressure**
The average of the maximum MVP in each measurement for each mask.

• **Contact areas**
The mean value of the maximum contact areas in each mask.

• **Contact time**
The average contact time (measured in ms) for each mask in each measurement.

• **Contact time [\%ROP]**
The average contact time (measured in \%ROP) for each mask in each measurement.

• **Begin of contact [\%ROP]**
The average time (in \%ROP) the mask begins loading.

• **End of contact [\%ROP]**
The average time (in \%ROP) the mask ends loading.

• **Pressure-time integrals**
The mean value of the pressure time integrals for each mask. The pressure-time integral corresponds to the area under the peak pressure curve.

• **Pressure-time integrals above threshold**
The mean pressure time integral above a set threshold.

• **Force-time integrals**
The mean value of the force time integral for each mask. The force-time integral represents the area under the force curve.

• **Force-time integrals above threshold**
The mean force time integral above a set threshold.

### 21.2.2 Time Processes

These are four time process variables calculated:

- Time processes of area
- Time processes of force
- Time processes of peak pressure
- Time processes of mean pressure

These windows calculate, at each frame, the mean value over all data files for each parameter (loaded area, force, peak and mean pressure). These values are shown vs. time. The two additional curves around the parameter define the 95% confidence interval.
21.3 Options for Displaying the Results

With the Pictures menu or the button in the toolbar the selection of displayed pictures can be changed.

With Options the scales for the graphs can be selected.
1. Select the parameter scale to change with a checkmark
2. Set the desired maximum value of the scale.
3. Refresh the parameter window (maximize and restore the window is one way of doing this).

Use the View menu to switch between standard deviation (Deviation) and confidence intervals.

Use the Window menu to arrange curves and diagrams on the screen.

21.4 Printing and Saving of Data

In the File menu various commands can be used to print and save data. Examples of the ASCII outputs are shown at the end of this section.

Print:
- **Print** – this option prints all parameters selected. A window will appear to select the desired parameters.
- **Print screen** – this option prints a copy of the screen.
- **Print Active Window** – this option prints only the selected window.
- **Print single values** – this option prints a table of all selected parameters in ASCII format.
- **Print statistics** – this option prints the statistical analysis in ASCII format.

Save:
- **Save single values** this option saves a table of all parameters in ASCII format (*.lst) where the numerical values for all files and all masks are listed.
- **Save statistics** this option saves the entire statistical analysis in ASCII format (*.sta). For each parameter and each mask (and the total object), minimum and maximum values, range, mean value, variance, standard deviation, coefficient of variation, number of files and the borders of the 95% confidence interval are listed.
- **Save table of all the processed values** - this option saves all parameters for each mask and each data file in ASCII format (*.rec). There are two options available: Old format and New format. In both formats the data and mask files are always listed in the first two columns. After that the parameters are listed for the whole object and the masks in the following order:
  - **old format**
    - Maximum contact area
    - Maximum force
    - Peak pressure
    - Contact time in ms
    - Contact time in % ROP
    - Begin of contact in % ROP
    - End of contact in % ROP
- Pressure-Time Integral
- Force-Time Integral
- Moment of peak pressure in % ROP
- Moment of maximum force in % ROP

- **new format**
  - Maximum contact area
  - Maximum force
  - Peak pressure
  - Maximum mean pressure
  - Contact time in ms
  - Contact time in % ROP
  - Begin of contact in % ROP
  - End of contact in % ROP
  - Pressure-time Integral
  - Force-time Integral
  - Instant of peak pressure in ms
  - Instant of peak pressure in % ROP
  - Instant of maximum force in ms
  - Instant of maximum force in % ROP
  - Pressure-time Integral above threshold
  - Force-time Integral above threshold
  - Mean force
  - Mean contact area

- **Save averaged time processes** - this option saves the time dependence of the mean values of loaded area, force, peak pressure and mean pressure in ASCII format (*.hin) for the total object and each mask.

**Additional functions:**
- These options allow the calculations of the program to be saved and opened without having to recalculate the parameters.
- **Save results of calculation to binary file** - this option saves the results of the evaluation.
- **Load results of calculation from binary file** - this option loads the saved results of an evaluation.

**Example ASCII outputs:**

**Save single values**

groupmask evaluation
novel electronics, inc.
pedar is a registered trademark of novel

Group : c:\nove\nove\win\par\123456.par
Number of members : 0

Declaration of masks
1. Contact areas [cm²]

<table>
<thead>
<tr>
<th>DAT</th>
<th>MASK</th>
<th>Total</th>
<th>M01</th>
<th>M02</th>
<th>M03</th>
<th>M04</th>
<th>M05</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.dat</td>
<td>1.msa</td>
<td>131.25</td>
<td>32.25</td>
<td>13.75</td>
<td>54.5</td>
<td>13.5</td>
<td>4.5</td>
</tr>
<tr>
<td>2.dat</td>
<td>2.msa</td>
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**Save statistics**

groupmask evaluation
novel electronics, inc.
pedar is a registered trademark of novel

Group file: c:novel\novelwin\part123456.par

Number of members: 6

Declaration of variables

| V01: Contact areas [cm²] |
| V02: Maximum force [N] |
| V03: Peak pressure [N/cm²] |
| V04: Contact time [ms] |
| V05: Contact time [%ROP] |
| V06: Begin of contact [%ROP] |
| V07: End of contact [%ROP] |
| V08: Pressure time [N/cm²]*s |
| V09: Force time [(N)*s] |
| V10: Instant of peak pressure [ms] |
| V11: Instant of peak pressure [%ROP] |
| V12: Instant of [ms] |
max, **force**

V13: force [%ROP]

Pressure time [(N/cm²)*s]

V14: integrals above threshold 0.00 N/cm²

Force-time Integrals [(N)*s] above threshold 0.00 N/cm²

V15: Maximum mean

V16: pressure [N/cm²]

Mean

V17: force [N]

V18: Mean area [cm²]

Range: maximum-minimum

Mean Value: \[ \text{sum}(x[i])/n; i=1,n (n=number of members) \]

Variance: \[ \text{sum}(x[i]-\text{mean value})^2)/(n-1) \]

Standard Deviation: square root(variance)

100*standard deviation/

Variation Coefficient: mean value [%]

95% Confidence Interval: mean value ± standard deviation * t(prob=0.95,n-1)/square root(n)

Declaration of masks

M01: mask_01
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M03: mask_03
M04: mask_04
M05: mask_05
M06: mask_06

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